

Instrument for the treatment of skin zones

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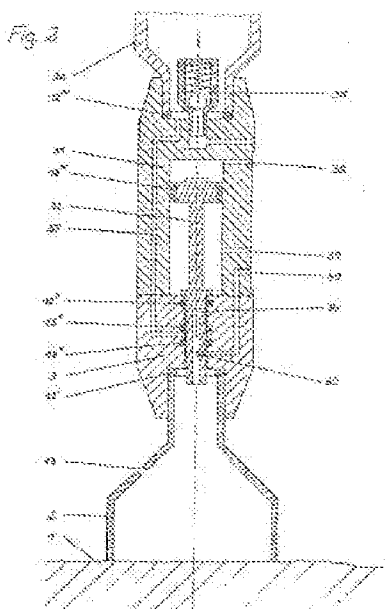
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 EP0784997 (A3)

Abstract of AT 400305 (B)

This invention relates essentially to an instrument for the treatment of specific skin zones by means of various therapeutic measures in the form of an appliance, characterized in that there is provided a bell-shaped vessel 2 which is capable of being placed sealingly with its free open end onto the skin zone 1 to be treated and into the opposite end of which are inserted or are capable of being inserted units 3, 3a having separate outlets and intended for the application of a vacuum, coloured light and preferably atomized active substances, the application units 3, 3a being releasably connected or individually connectable via supply lines separate from one another, specifically a vacuum hose 10 and two control lines for compressed air, serving for generating and regulating the active substance mist and designed as hose lines 11 and 11a, or a vacuum hose 10 and a light guide 7; to three independent supply units, specifically an instrument 4 for the regulatable generation of a vacuum, a coloured-light source 5 emitting selectable colours and an active-substance control unit 6 feeding compressed air to the hose lines, so that the individual therapeutic measures (vacuum, coloured-light and active-substance application) can be used either simultaneously or separately or in any desired combination, and in that, furthermore, other bell-shaped vessels of various types and sizes, capable of being connected to the application units 3 or 3a, are provided.



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PATENTCHRIFT

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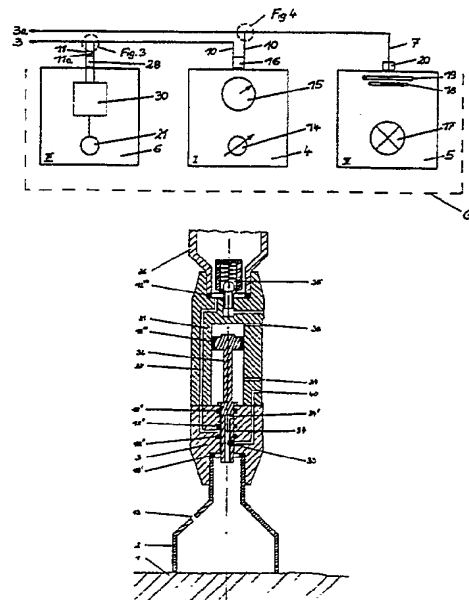
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(54) EINRICHTUNG ZUR BEHANDLUNG VON HAUTZONEN

(57) Im wesentlichen handelt es sich bei dieser Erfindung um eine Einrichtung zur Behandlung bestimmter Hautzonen mittels verschiedener Therapiemaßnahmen in Form eines Gerätes, dadurch gekennzeichnet, daß ein mit seinem freien, offenen Ende auf die zu behandelnde Hautzone (1) dicht aufsetzbares, glockenförmiges Gefäß (2) vorgesehen ist, in dessen gegenüberliegenden Ende, getrennte Ausgänge aufweisende Einheiten (3, 3a) zur Applikation von Unterdruck, Farblicht und vorzugsweise vernebelten Wirkstoffen eingesetzt bzw. einsetzbar sind, wobei die Applikationseinheiten (3, 3a) über voneinander getrennte Versorgungsleitungen und zwar einem Vakuumschlauch (10) und zwei zum Erzeugen und Regeln des Wirkstoffnebels dienenden, als Schlauchleitungen (11 und 11a) ausgebildete Steuerleitungen für Druckluft beziehungsweise einem Vakuumschlauch (10) und einem Lichtleiter (7) mit drei unabhängigen Versorgungseinheiten und zwar eine Einrichtung (4) zur regelbaren Erzeugung von Unterdruck, einer wählbaren Farben abgebenden Farblichtquelle (5) und einer die Schlauchleitungen mit Druckluft speisenden Wirkstoffsteuereinheit (6), lösbar verbunden bzw. einzeln verbindbar sind, sodaß die einzelnen Therapiemaßnahmen (Unterdruck-, Farblicht- und Wirkstoffapplikation) entweder gleichzeitig oder getrennt, oder beliebig kombiniert anwendbar sind und daß ferner weitere mit den Applikationseinheiten (3 oder 3a) verbindbare, glockenförmige Gefäße verschiedener Bauart und Größe vorgesehen sind.



Die Erfindung betrifft eine Einrichtung zur Behandlung bestimmter Hautzonen mittels verschiedener Therapiemaßnahmen in Form eines Gerätes.

Im besonderen handelt es sich bei diesen Therapiemaßnahmen um die Applikation von Unterdruck, Farblight und Wirkstoffen auf die zu behandelnde Hautzone.

- 5 Diese Therapieformen, bzw. die dafür verwendbaren Einrichtungen, sind einzeln bereits lange Zeit in Anwendung, wie ein dicht auf die zu behandelnde Körperoberfläche aufsetzbares Gefäß, ab jetzt als Glocke bezeichnet, welches mit einer Einrichtung zur Erzeugung von Unterdruck verbunden ist und unter dem Ausdruck "Schröpfungsglas" allgemein bekannt ist.

- Des weiteren ist z.B. aus der DE-A1-3301802 bekannt, in diesen Glocken Lichtquellen mit optischen Gittern und Filtern zu Bestrahlungszwecken, zu montieren.

- 10 Dadurch entstehen jedoch konstruktionsbedingt Nachteile, wie z.B. die Unhandlichkeit der Glocke durch große Außenabmessungen, da alle Komponenten darin eingebaut sind, oder bei einer konstruktiv aufwendigen Verkleinerung der Glocke kommt es durch die schlechte Wärmeabfuhr zur Temperaturerhöhung und dadurch zu thermischen Problemen an der zu bestrahlenden Oberfläche; als Beispiel sei blaues Licht
15 angeführt, welches ein kaltes Licht ist und, um seine volle therapeutische Wirkung zu entfalten, nicht mit einer Temperaturerhöhung an der bestrahlten Oberfläche verbunden sein darf.

- Die vorliegende Erfindung löst dieses Problem dadurch, daß ein mit seinem freien, offenen Ende auf die zu behandelnde Hautzone (1) dicht aufsetzbares, glockenförmiges Gefäß (2) vorgesehen ist, in dessen gegenüberliegenden Ende, getrennte Ausgänge aufweisende Einheiten (3, 3a) zur Applikation von Unterdruck, Farblight und vorzugsweise vernebelten Wirkstoffen eingesetzt bzw. einsetzbar ist, wobei die
20 Applikationseinheiten (3 bzw. 3a) über voneinander getrennte Versorgungsleitungen und zwar einen Vakuumschlauch (20) und zwei zum Erzeugen und Regeln des Wirkstoffnebels dienenden Schlauchleitungen, ausgebildete Steuerleitungen für Druckluft (11 u. 11a), beziehungsweise einem Vakuumschlauch (10) und einem Lichtleiter (7) mit drei unabhängigen Versorgungseinheiten und zwar eine Einrichtung (4) zur
25 regelbaren Erzeugung von Unterdruck, einer wählbaren Farben abgebenden Farblightquelle (5) und einer, die Schlauchleitungen mit Druckluft speisenden Wirkstoffsteuereinheit (6), lösbar verbunden bzw. einzeln verbindbar sind, sodaß die einzelnen Therapiemaßnahmen (Unterdruck -, Farblight- und Wirkstoffapplikation) entweder gleichzeitig oder getrennt, oder beliebig kombiniert anwendbar sind und daß ferner weitere mit den Applikationseinheiten (3 u. 3a) verbindbare, glockenförmige Gefäße verschiedener Bauart und Größe
30 vorgesehen sind.

Durch die Verwendung von Lichtleitern wird nur das Licht selbst übertragen, aber keine Wärme. Weiters ermöglicht diese Konstruktion es, die glockenförmigen Gefäße in den Abmessungen klein zu halten und vom Anwender gegen andere Gefäße selbständig austauschen zu lassen.

- Die erfindungsgemäße Einrichtung unterscheidet sich im Bezug auf die Farblight-Therapie weiters von
35 bisherigen Entwicklungen (DD-A1-219676) dadurch, daß nur Licht bestimmter Wellenlängen übertragen wird. Um die erwünschte Wirkung bei der therapeutischen Anwendung zu erzielen, sind Lichtquellen im Bereich der Spektralfarben möglichst ohne störenden UV-Anteil erforderlich, zu diesem Zweck werden, um die erwünschte Wirkung zu erzielen, erfindungsgemäß spezielle Farbfilterscheiben verwendet und der unerwünschte UV-Anteil von einem weiteren Filter zum größtmöglichen Teil absorbiert.

- 40 Bei Einrichtungen zur Wirkstofftherapie sind verschiedene Einrichtungen bereits bekannt, wie z.B. das Verdampfen von Wirkstoffen, welche eingeatmet werden. Die Erfindung liegt in diesem Fall auch darin, daß durch die Kombination der Unterdruck - und Wirkstoffapplikation in der Glocke, eine weit größere Wirksamkeit auf der zu behandelnden Körperoberfläche entfaltet wird, wobei noch zusätzlich die Farblight-Therapie eingesetzt werden kann.

- 45 Daher kann gesagt werden, daß mit einer Kombination dieser Therapiemaßnahmen gemäß der vorliegenden Erfindung sich starke Wirkungen hinsichtlich der Erzeugung von Reizen und anderen Vorgängen in bestimmten Hautzonen des Menschen erzielen lassen.

Die Wirkung der Therapie beschränkt sich jedoch nicht nur auf Bindegewebsschichten, Blut und Lymphe, sondern erregt die Nerven in der Haut und die darunter liegenden Weichteile wie z.B. Muskeln und Sehnen.

- 50 Sie wirkt auf tiefliegende Nervenstämmen und bringt dadurch vielfältige sich auf die inneren Organe übertragende Reize hervor.

- Die Therapie in ihren verschiedenen Formen bedingt reaktive Hyperämie, fördert den Zufluß und Abfluß der Gewebssäfte, erhöht die Permeabilität der Kapillarwände, tonisiert innere Organe, fördert die Ausscheidung, unterstützt die Blutzirkulation bei insuffizienten Kreislauf, übt Rückwirkung auf das allgemeine
55 Kraftgefühl, auf Ermüdungszustände, Stimmung und Psyche aus.

Ein Ausführungsbeispiel des Erfindungsgegenstandes wird nachfolgend in den Figuren 1 - 6 erläutert.

Die Figur 1 zeigt die erfindungsgemäße Einrichtung zur kombinierten Unterdruck-, Farblight- und Wirkstofftherapie ohne glockenförmiges Gefäß (2), Applikationseinheiten (3 bzw. 3a), wobei die drei Versor-

gungseinheiten (4,5,6) je nach Ausführung in ein gemeinsames Gehäuse (G), wie auch in separaten Modulen (I,II,III) eingebaut sein können.

In Figur 2 wird das auf die zu behandelnde Hautzone (1) dicht aufsetzbare und somit mit seinem Innenraum hermetisch abschließbare glockenförmige Gefäß (2), nachfolgend Glocke bezeichnet, gezeigt, wobei die Glocke (2) mit der Unterdruck- und Wirkstoffapplikations-Einheit (3) verbunden bzw. 5 verbindbar ist.

Die Applikations-Einheiten (3 u. 3a) sind wiederum mit der Einrichtung zur Erzeugung von Unterdruck (4) der Farblicht-Quelle (5) und der Wirkstoffsteuer-Einheit (6) durch mit für die einzelnen Medien geeignete Versorgungsleitungen verbunden.

Die Versorgungsleitungen sind als Lichtleiter (7) und als drei weitere Schlauchleitungen (10, 11 und 11a) ausgebildet. 10

Die Figur 3 zeigt eine mögliche Ausführungsform der Unterdruck u. Wirkstoffapplikationsleitung, wobei der Vakuumschlauch (10) als Ummantelung für die anderen Leitungen (11 u. 11a) ausgeführt ist.

Die Figur 4 zeigt eine mögliche Ausführungsform der Unterdruck u. Farblicht -Therapieleitung, wobei der 15 Vakuumschlauch (10) als Ummantelung für den Lichtleiter (7) ausgeführt ist. Die einzelnen Lichtleiterfasern (7) können dann lose im verbleibenden Raum zwischen Vakuumschlauch (10) geführt werden, wobei bei der Aufteilung der Versorgungsleitungen zu den einzelnen Versorgungseinheiten (4,5 u.6) dann eine Dichtung (12) zur Aufrechterhaltung des Unterdruckes vorgesehen ist.

Aus diesem Grund sind auch mehrere Dichtungen (12' u. 12'') zwischen Glocke (2) und Applikations- 20 Einheit (3 bzw. 3a) sowie in der Applikationseinheit (3) vorgesehen.

Da Unterdruckabbau aber zur leichteren Handhabung der Glocke für therapeutische Zwecke notwendig ist, ist die Glocke (2) mit einer Öffnung (13) versehen, welche von Hand geschlossen werden kann.

In Figur 5 wird die Einmündung der Lichtleiterfasern (7) in die Applikationseinheit (3a) verdeutlicht. Die Lichtleiterfasern (7) werden in eine Hülse (23) eingepreßt und somit fixiert und gebündelt, wobei der 25 Unterdruck in der verbleibenden Fläche zwischen den einzelnen Lichtleiterfasern ungehindert aufrechterhalten werden kann. Die Hülse (23) wird mit einem Ende mit der Applikationseinheit (3a) verbunden; über das andere Ende wird der Vakuumschlauch (10) geschoben, wodurch die Dichtheit des Systems gewährleistet wird.

In Figur 6 wird die Verbindung der einzelnen Leitungen, bei der Unterdruck und Wirkstoffapplikation, mit 30 der Applikationseinheit (3) verdeutlicht. Im wesentlichen handelt es sich hierbei um einen Verteilerblock (24), welcher die Verbindungen zwischen Schlauchleitung (11) und einem Steuerluftkanal (38) der Applikationseinheit (3), beziehungsweise Schlauchleitung (11a) und einem zweiten Steuerluftkanal (39) der Applikationseinheit (3), beziehungsweise Vakuumschlauch (10) und einem Vakuumkanal (40) der Applikationseinheit (3) herstellt.

Der Unterdruck in der Glocke (2) wird mittels der regelbaren Einrichtung (4) bestimmt, wobei eine mögliche Ausführungsform ein von Hand zu verstellendes Drosselventil (14) mit einer visuellen Anzeigeein- 35 heit (15) ist. Die Verbindung (16) zur Transportleitung (10) kann lösbar ausgeführt sein.

In der Farblicht-Quelle (5) ist eine Lichtquelle (17) installiert, über UV-Sperrfilter (18) und FarbfILTER (19) gelangt das Licht in den mittels lösbarer Verbindung (20) befestigten Lichtleiter (7), der zur Glocke (2) führt.

In der Wirkstoffsteuer-Einheit (6) wird Druckluft geregelt, welche für den Vernebelungsvorgang der Wirkstoffe (ätherische Öle) in der Aufbereitungseinheit (Fig. 2) notwendig ist. Der dafür nötige Druck wird durch eine Pumpe (21), die sich in der Wirkstoffsteuer-Einheit (6) befindet, erzeugt. Mittels lösbarer 40 Anschluß (28) und der als Schlauch ausgeführten Wirkstoffsteuer-Leitungen (11 u. 11a) wird wiederum die Verbindung zur Applikationseinheit (3) hergestellt.

45 Funktionsbeschreibung

Um den gewünschten therapeutischen Effekt zu erzielen, muß der Unterdruck in der Glocke (2), mittels des Drosselventiles (14) in der Einrichtung zur Unterdruckerzeugung (4), an der jeweiligen Hautoberfläche 50 (1) angepaßt werden. Durch Öffnen oder Verschließen der Öffnung (13) an der Glocke (2) kann der Anwender die Dauer des an der Haut (1) anliegenden Unterdruckes bestimmen.

Das Farblicht wirkt ununterbrochen auf die Hautoberfläche in der gewünschten Farbe (Wellenlänge) ein.

In der Applikationseinheit (3) wird der Wirkstoff vernebelt. Nach dem Betätigen eines Tasters, steuert ein Magnetventil (30) die von der Pumpe (21) erzeugte Druckluft, welche in weiterer Folge durch die 55 Wirkstoffsteuerleitungen (11 u. 11a) in das Innere eines Pneumatikzylinders (31) der Applikationseinheit (3) geleitet wird. Ein Wirkstoffsteuerkolben (32) mit seinem Kolbenring (12''') bewegt sich nun für einen kurzen Augenblick nach oben; in diesem Moment ist ein im Wirkstoffsteuerkolben (32) axial verlaufender Druckluftkanal (34) über eine obere radial verlaufende Durchgangsbohrung (34') mit dem Inneren des Pneumatikzy-

linders (31) verbunden und führt somit Druckluft.

Die Düsenluft strömt nun entlang des Druckluft-Kanales (34) bis zu einer Düse (33), welche quer in den Druckluftkanal (34) einmündet und in diesem Moment mit einem durch die Wand des Pneumatikzylinders (31) verlaufenden Wirkstoffkanal (37) verbunden ist.

Durch die vorbeiströmende Luft entsteht eine Sogwirkung in der Düse (33), wodurch der Wirkstoff in den Luftstrom gelangt und vernebelt wird. Der Wirkstoffkanal (37) leitet den Wirkstoff von einem Wirkstoffbehälter (36) zur Applikationseinheit (3). Um den im Wirkstoffbehälter (36) auftretenden Unterdruck zu kompensieren befindet sich in diesem ein Lüftungsventil (35); zwischen den auf die Applikationseinheit (3) aufsetzbaren Wirkstoffbehälter (36) und der Applikationseinheit (3) befindet sich ebenfalls eine Ringdichtung (12''').

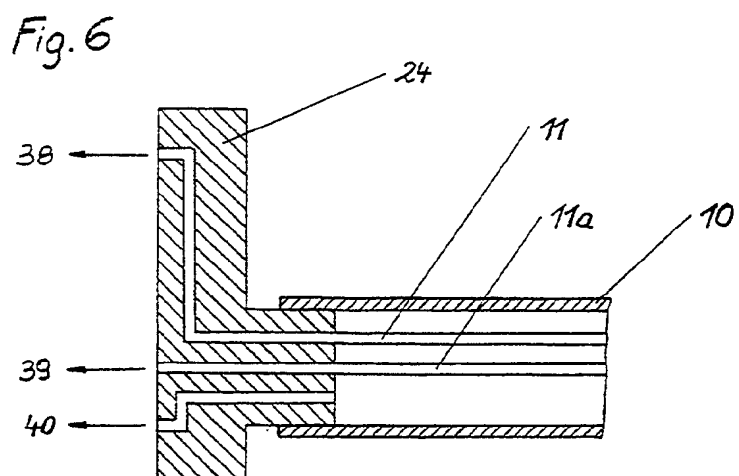
Dieser Vernebelungsvorgang wird entweder durch einen Fußtaster, oder durch einen Taster, welcher sich an der Wirkstoffsteuereinheit befindet ausgelöst. Die Dauer der Vernebelung wird durch eine in der Wirkstoffsteuereinheit integrierten Elektronik geregelt, unabhängig davon, wie lange man einen dieser beiden Taster auslöst. Mittels eines Reglers an der Wirkstoffsteuereinheit kann die Länge des Vernebelungsvorganges, an die verschiedenen voluminösen Glocken und an die unterschiedlichen Unterdrücke der einzelnen Behandlungen so angepaßt werden, damit gewährleistet ist, daß in den Glocken immer ein Unterdruck besteht.

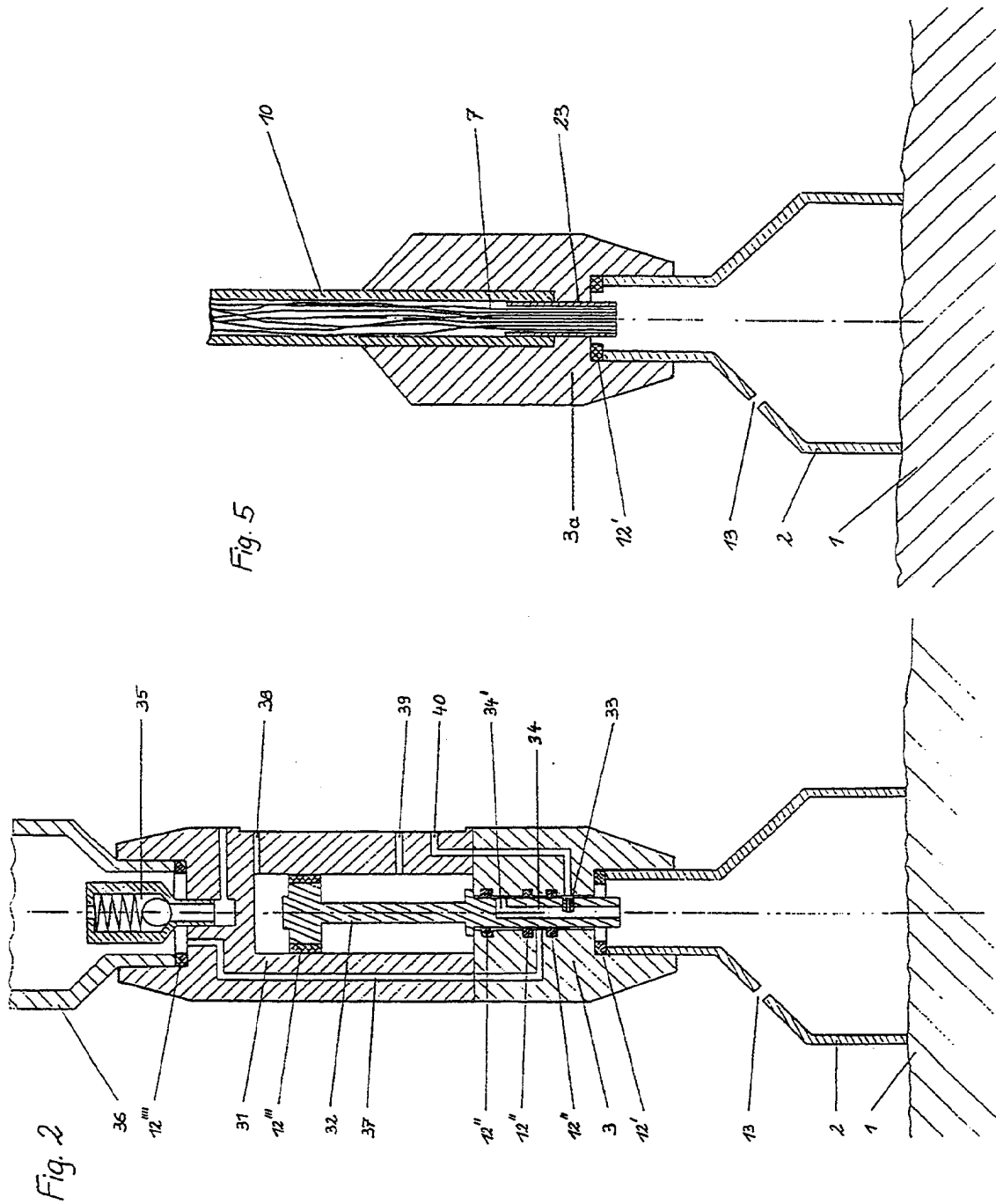
Die beschriebene Konstruktion der beliebigen Anschließbarkeit der Versorgungseinheiten (4,5 u.6) ermöglicht es den Anwendern auch die einzelnen Einheiten (4,5 u.6) jeweils alleine zu Betreiben, wie auch wahlweise miteinander zu kombinieren, oder sie gleichzeitig zu betreiben.

Patentansprüche

1. Einrichtung zur Behandlung bestimmter Hautzonen mittels verschiedener Therapiemaßnahmen in Form eines Gerätes, **dadurch gekennzeichnet**, daß ein mit seinem freien, offenen Ende auf die zu behandelnde Hautzone (1) dicht aufsetzbares, glockenförmiges Gefäß (2) vorgesehen ist, in dessen gegenüberliegenden Ende, getrennte Ausgänge aufweisende Einheiten (3, 3a) zur Applikation von Unterdruck, Farblicht und vorzugsweise vernebelten Wirkstoffen eingesetzt bzw. einsetzbar ist, wobei die Applikationseinheiten (3, 3a) über voneinander getrennte Versorgungsleitungen und zwar einem Vakuumschlauch (10) und zwei zum Erzeugen und Regeln des Wirkstoffnebels dienenden, als Schlauchleitungen (11 und 11a) ausgebildete Steuerleitungen für Druckluft beziehungsweise einem Vakuumschlauch (10) und einem Lichtleiter (7) mit drei unabhängigen Versorgungseinheiten und zwar einer Einrichtung (4) zur regelbaren Erzeugung von Unterdruck, einer wählbaren Farben abgebenden Farblichtquelle (5) und einer die Schlauchleitungen (11 und 11a) mit Druckluft speisenden Wirkstoffsteuereinheit (6), lösbar verbunden bzw. einzeln verbindbar sind, sodaß die einzelnen Therapiemaßnahmen (Unterdruck-, Farblicht- und Wirkstoffapplikation) entweder gleichzeitig oder getrennt, oder beliebig kombiniert anwendbar sind und daß ferner weitere mit den Applikationseinheiten (3, 3a) verbindbare, glockenförmige Gefäße verschiedener Bauart und Größe vorgesehen sind.
2. Einrichtung nach Anspruch 1, **dadurch gekennzeichnet**, daß das glockenförmige Gefäß (2) mit einer vom Anwender verschließbaren Öffnung (13) versehen ist.
3. Einrichtung nach Anspruch 1 oder 2, **dadurch gekennzeichnet**, daß die Farblichtquelle (5) neben einer Lichtquelle (17) und einer Farbfilterscheibe (18) einen UV-Sperrfilter (19) aufweist.
4. Einrichtung nach Anspruch 1, 2 oder 3, **dadurch gekennzeichnet**, daß in der Wirkstoffapplikationseinheit (3) Mittel (32,34,12'') vorgesehen sind, welche nur während der durch Druckluftzufuhr erfolgenden Vernebelung des Wirkstoffes die Steuerleitungen (11 u. 11a) für Druckluft mit dem Innenraum des Glockenförmigen Gefäßes (2) verbinden, um den im Inneren der Glocke (2) herrschenden Unterdruck im wesentlichen aufrecht zu erhalten.
5. Einrichtung nach Anspruch 1,2,3 oder 4, **dadurch gekennzeichnet**, daß ein Wirkstoffbehälter (36) direkt auf die Wirkstoffapplikationseinheit (3) dicht aufsetzbar und mittels eines Wirkstoffkanales (37) mit der Vernebelungseinheit verbunden ist.

Hiezu 2 Blatt Zeichnungen







(12) AUSTRALIAN PATENT ABRIDGMENT
(19) AU

(11) AU-A-18515/83

(54) LAMP FOR IRRADIATION OF TUMOURS
(71) THE UNIVERSITY OF ADELAIDE, FREDERICK JOHN JACKA AND
ALISTAIR JOSEPH BLAKE
(21) 18515/83 (22) 27.8.82
(23) 29.8.83 (24) 27.8.82
(43) 1.3.84 (44) 31.7.86
(51)³ A61N 5/06 F21V 9/12 F21V 7/09 F21V 13/08
(74) CO
(56) US 4068931
GB 1325838
FR 613587

(57) Claim

1. Apparatus for effecting useful irradiation of tumours having therein a photoactive drug, including an incandescent filament to provide a light source, a reflector surrounding the incandescent filament a characteristic of which reflector with respect to an outlet aperture within the apparatus is that light when emanating from the filament will be at least substantially directed and reflected to pass through the aperture so as not to have greater than a selected angle of diversion from an axis of a lamp, a chamber connected to the outlet aperture being otherwise shaped so that light passing therethrough from the filament will retain at least substantially its entry angle of divergence with respect to the axis of the lamp and containing therewithin a liquid effective to provide absorption of a range of wave lengths of light passing therethrough, and at an end of said chamber filters including an interference filter positioned and shaped and of such a size that light emanating from the said filament and having passed through the said liquid will strike said interference filter with an angle of incidence to the filter not greater than a selected suitable value.

COMPLETE

SPECIFICATION

(ORIGINAL)



FOR OFFICE USE:

Class

Int. Class

Application Number :
Lodged :

18515/83

Complete Application No. :
Specification Lodged :
Published :

Priority:

Related Art:

This document contains the
amendments made under
section 49.

and is correct for printing.

TO BE COMPLETED BY APPLICANT

Name of Applicant s

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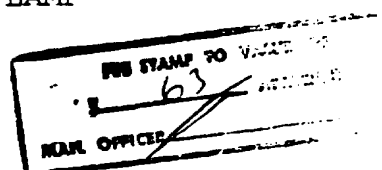
Complete Specification for the invention entitled:

LODGED AT SUB-OFFICE

29 AUG 1983

Adelaide

"AN IMPROVED LAMP"

The following statement is a full description of this invention, including the best method of performing
it known to us.

This invention relates to an improved lamp especially intended for use for irradiation of malignant tumours where these have had incorporated in them a receptive photoactive drug.

5. It is known for instance that "HAEMATOPORPHYRIN DERIVATIVE" when injected into the human body or into other living animals can have the effect of selectively making tumours more susceptible to destruction when irradiated with a selected radiation than healthy tissue.

10. There are considerable difficulties surrounding the use of this technique.

It is desirable to have a very high intensity of irradiation onto the affected tissue but it can be a significant problem that if the radiation is too intense especially with the healthy tissue, significant tissue damage can result.

15.

It is known to propose a laser for use in this application in that certain laser beams can be generated in a very narrow bandwidth.

The problem however with lasers is that it is very difficult to obtain a high power of transmission and the result of this to get sufficient radiation in respect of the tissue is that the irradiation must take a substantial period of time.

20.

Considering some applications such as the case where the patient has been subject to surgery with the vulnerable areas open, it is preferable that the patient should not

25.

Perhaps more importantly, by using an incandescent light source, a sufficient output within the necessarily restricted range of wave lengths can be achieved to significantly lessen the time of irradiation needed as compared to previous means available.

Furthermore, it has been found that by careful selection of design criteria, an incandescent lamp for this purpose can be achieved which has small outer dimensions so that the lamp can be substantially rugged and portable, and both economic and useful in a variety of applications.

A 5.

to the axis of the lamp and containing therewithin a liquid effective to provide absorption of a range of wave lengths of light passing therethrough, and at an end of said chamber, filters including an interference filter positioned and shaped and of such a size that light resulting from the said filament and having passed through the said liquid will strike said interference filter with an angle of incidence to the filter not greater than a selected suitable value.

5.

Preferably a reflector is located so as to accept light passing through the several filters and so reflect this so as to concentrate the light through a smaller outlet there being as a consequence a larger range of angles of divergence through which the light will be distributed.

10.

Preferably, the maximum angle of divergence which is selected for the light emanating from the aperture of the reflector surrounding the filament is approximately 30 degrees.

15.

Preferably the liquid within the chamber providing for selective light absorption is water and in a preferable arrangement, the water which will absorb light in the infrared range of wave lengths is also used to cool the reflector and is recirculated.

20.

Preferably, the filament providing for the incandescent light output is substantially coiled into a cylindrical shape and is coaxial with the axis of the reflector and the lamp.

25.

It has been discovered that by providing a compound paraboloidal reflector associated with a filament positioned as described, that this provides an effective reflector

which can occupy a reasonably small space and which can nonetheless provide light available from the filament emanating from an exit aperture within a selected limited angle of divergence from the axis of the lamp.

5. To assist in an understanding of the invention, this will be described with respect to a particular apparatus which shall be described with the assistance of drawings attached hereto.

10. In Fig. 1 of the attached drawings this shows in cross-section the preferred embodiment and Fig. 2 is a schematic diagram of the end concentrator and Fig. 3 is a schematic diagram showing the specific required relationships between the surface of the reflector, the exit aperture, the angle of maximum divergence from the axis of the lamp
15. and the filament as used in the preferred embodiment.

Referring in the first instance to Fig. 3, there is a filament that can be assumed to be of cylindrical shape of radius (r_F) length (l) aligned along an axis (OX) and having a front edge at (FF').

20. The exit aperture of the reflector is defined by (AA') and the portion (A) to (B) of the cross-section has a shape which is parabolic having a focus at (F') and axis parallel to (FA') making an angle ω_0 equal to the maximum angle of divergence from the axis of the lamp. The portion (B) to (C)
25. is a parabola having a focus at (F) and axis (FA').

The portion (C) to (D) is of circular shape having a centre at (F).

- The location (C) is a linear extension of the line (A') to (F). The dimension (DD') is determined by the
30. diameter of the bulb (2 of Fig. 1)

The reflector has a central axis (OX) and is a surface of revolution accordingly.

5. Selection of an appropriate filament and having this located in relation to a selected outlet aperture of selected radius (r) determines with the maximum angle of divergence from the axis of the lamp the shape and size of the parabolic shape of the portion of (A) to (B) and likewise the shape of the parabolic portion (B) to (C) and the radius of the portion (C) to (D).
10. For given filament dimensions the fraction of light lost through the aperture DD' is determined by the distance ST. For particular filament dimensions and choice of ω_0 the radius (r) determines the distance ST.
15. The angle of maximum divergence from the axis of the lamp for light output shown as ω_0 has been selected in a practical situation as being limited to approximately 30 degrees and in fact 30 degrees in the preferred embodiment.
20. Referring now to Figure 1, this shows in cross-section a reflector 1 encircling an incandescent lamp with filament 3.
25. The outlet aperture shown as a window at 4 has a peripheral seal 5 and connects to a cylindrical reflector.
30. The cylindrical reflector 6 has an internal surface adapted to provide for reflection of the output signal from the reflector 1 and direct output from the filament 3.
30. At the outer end of reflector 6 are two filters a first at 7 being peripherally sealed at 8 and being of absorption type selected to provide a sharp cut off of absorption of any light below a selected wave length but in this particular case having regard to the tissue absorption any wave length less than 620nm.

A second filter at 9 is an interference filter providing an effectively sharp cut off of radiation above a selected wave length and having regard to the particular characteristics of the photoactive drug used, in this instance this is selected at approximately 680nm.

With this preferred embodiment, the interior of the cylindrical reflector 6 is filled with water and there is a circulation passage way providing for passage of the water through chamber 11 which is behind the first reflector 1 whereby to provide cooling of the reflector in this location.

To provide for concentration of the radiation, there is now provided a further reflector at 12 the characteristic of which is selected so as to concentrate the light through a smaller output aperture at 13 with a consequent expansion of the maximum angle of divergence from the axis of the lamp.

The shape of the generator of the reflector concentrator is now described with the assistance of Fig. 2 in which the portion of the generator from (A) through to (B) is a parabola whose focus is at (C') and whose axis is parallel to (CA') making an angle ω_1 to the axis OX and the shape of the generator from (B) to (C) is a straight line making an angle (ω_2) with the axis OX. BC is tangential to AB and the maximum angle of divergence of radiation exiting from the aperture CC' is ($\omega_1 + 2\omega_2$). The axis of revolution of the reflector is (OX).

In this way, by appropriate selection of the dimensions and criteria of the concentration, the maximum angle of divergence of radiation exiting from the exit aperture (CC') can be kept not greater than a desired value being 60° in the preferred embodiment.

Having now described in brief terms the embodiment, the result of use of this preferred embodiment is such that by using a tungsten filament quartz halogen bulb as the source of one kW of radiant energy such a lamp can be made relatively economically and nonetheless can deliver high power from an apparatus which is able to be relatively portable.

Careful selection of the design criteria and selection of reflector surfaces and selection of an arrangement of parts including a water absorber for the absorption of radiation of wave lengths substantially greater than 680nm has resulted in the lamp providing a high efficiency in terms of delivered output power over a suitably narrow wave length band.

In experimental trials, it has been shown to provide a useful source of such radiation over the effective wave lengths.

In addition to the preferred embodiment described, it is permissible to add or even take away some portions without departing at least from the broader concept of the invention.

Accordingly an additional concentrating lens can be added at the outlet 13.

The shape of this lens can be based on the same principles as that of reflector 12 so reducing the aperture and hence increasing the maximum angle of divergence.

Alternative lenses with the purpose of redistributing the light intensity can be designed to attach at 13 or 14. In one instance this may have the purpose of producing an approximately uniform light intensity.

The illustration has selected for matters of convenience a range of angles of divergence which are found to be preferred.

5. It is to be stressed that some variation from the selected range of angles of divergence from the axis of the lamp can occur without departing from the substantial spirit of this invention. Furthermore, the location and slope of the cut off edge of multi layer dielectric edge filters of the type that are used in the preferred embodiment varies with the
10. selected angle of incidence and hence for different angles of incidence may require appropriate modifications in the reflector design.

15. This is not to say that the substantial design of the reflector should alter but that the detailed dimensions may vary in accord with the variation in filament shape or design or for a different output angle of divergence from the axis of the lamp.

It is a facet of the design that it is a non-imaging device.

THE CLAIMS DEFINING THE INVENTION ARE AS FOLLOWS:-

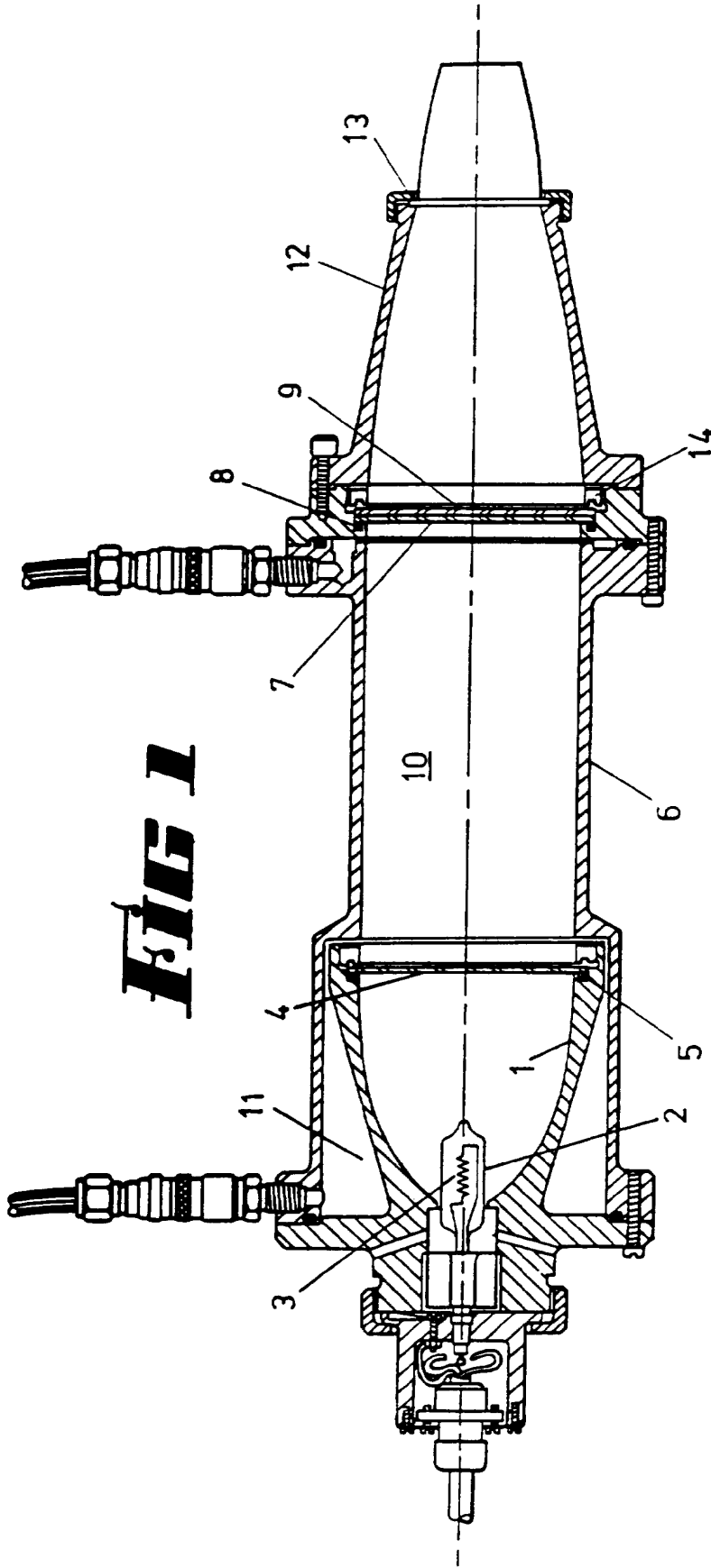
1. Apparatus for effecting useful irradiation of tumours having therein a photoactive drug, including an incandescent filament to provide a light source, a reflector surrounding the incandescent filament a characteristic of which
 5. reflector with respect to an outlet aperture within the apparatus is that light when emanating from the filament will be at least substantially directed and reflected to pass through the aperture so as not to have greater than a selected angle of diversion from
 10. an axis of a lamp, a chamber connected to the outlet aperture being otherwise shaped so that light passing therethrough from the filament will retain at least substantially its entry angle of divergence with respect to the axis of the lamp and containing therewithin a
 15. liquid effective to provide absorption of a range of wave lengths of light passing therethrough, and at an end of said chamber filters including an interference filter positioned and shaped and of such a size that light emanating from the said filament and having passed
 20. through the said liquid will strike said interference filter with an angle of incidence to the filter not greater than a selected suitable value.
-
2. An apparatus as in claim 1 further including a reflector shaped and located so as to accept light passing through the several filters including the interference filter at an end of said chamber, and adapted to reflect
 5. light passing through the said filters emanating from the filament so as to concentrate this light through a smaller outlet, there being as a consequence a range of angle of divergence through which the light will be distributed as it exists from the apparatus.

3. Apparatus as in either of claims 1 or 2 wherein the range or angle of divergence from the axis of the lamp as the light emanates from the aperture of the reflector surrounding the filament is approximately 30°.
- 5.
4. Apparatus as in any one of the preceding claims wherein the liquid within the chamber is water.
5. Apparatus according to any one of the preceding claims wherein the chamber is connected by conduit to heat exchange means whereby to provide for circulation of the liquid and cooling of the liquid thereby within the chamber.
- 5.
6. Apparatus according to any one of the preceding claims wherein the filament is of a substantially coiled cylindrical shape with the axis of the cylindrical shape being coaxial with the axis of the lamp.
7. Apparatus according to any one of the preceding claims wherein the reflector surrounding the filament is a compound paraboloidal reflector.
8. Apparatus according to any one of the preceding claims adapted to have a peak output of light of approximately 680 nm wave length.
9. Apparatus according to any one of the preceding claims which is adapted to most effectively irradiate tumours where the tumours contain the substance "HAEMATOPORPHYRIN DERIVATIVE".

10. Apparatus for effecting useful irradiation of tumours having therein a photoactive drug substantially as described in the specification with reference to and as illustrated by the accompanying drawings.

Dated this 29th day of May, 1986.

THE UNIVERSITY OF ADELAIDE,
FREDERICK JOHN JACKA and
JOSEPH BLAKE,
By their Patent Attorneys,
COLLISON & CO.



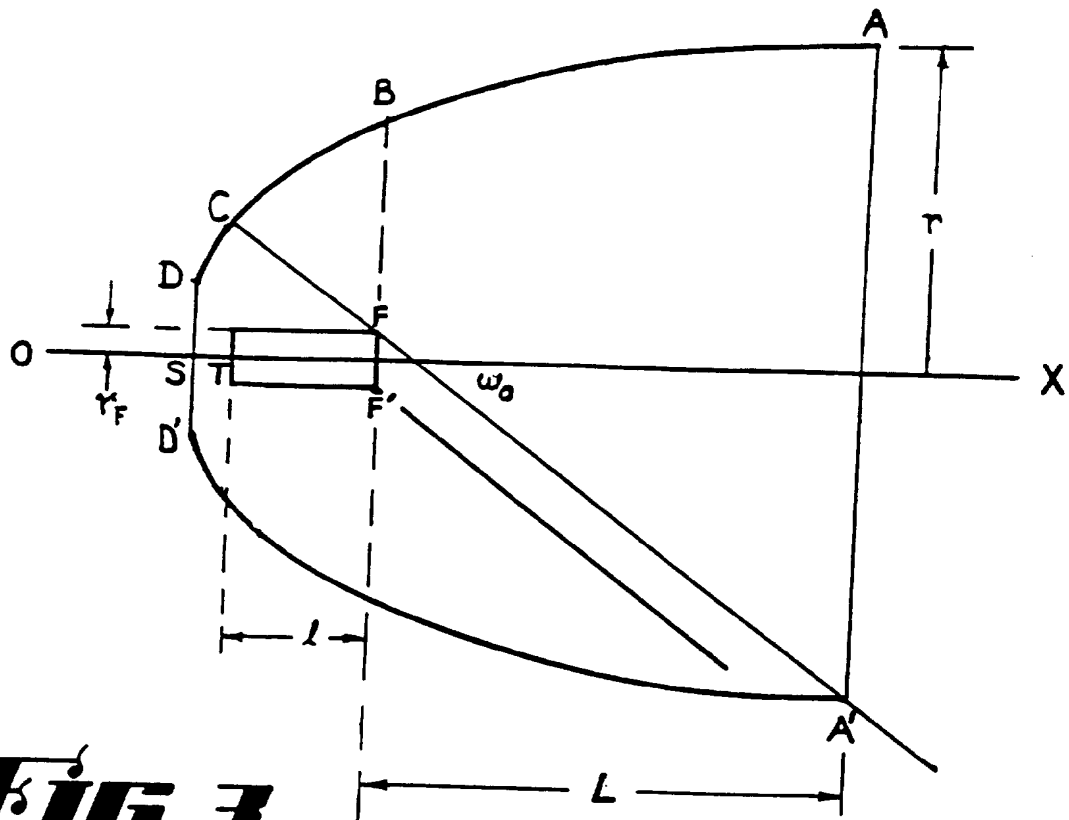


FIG 3

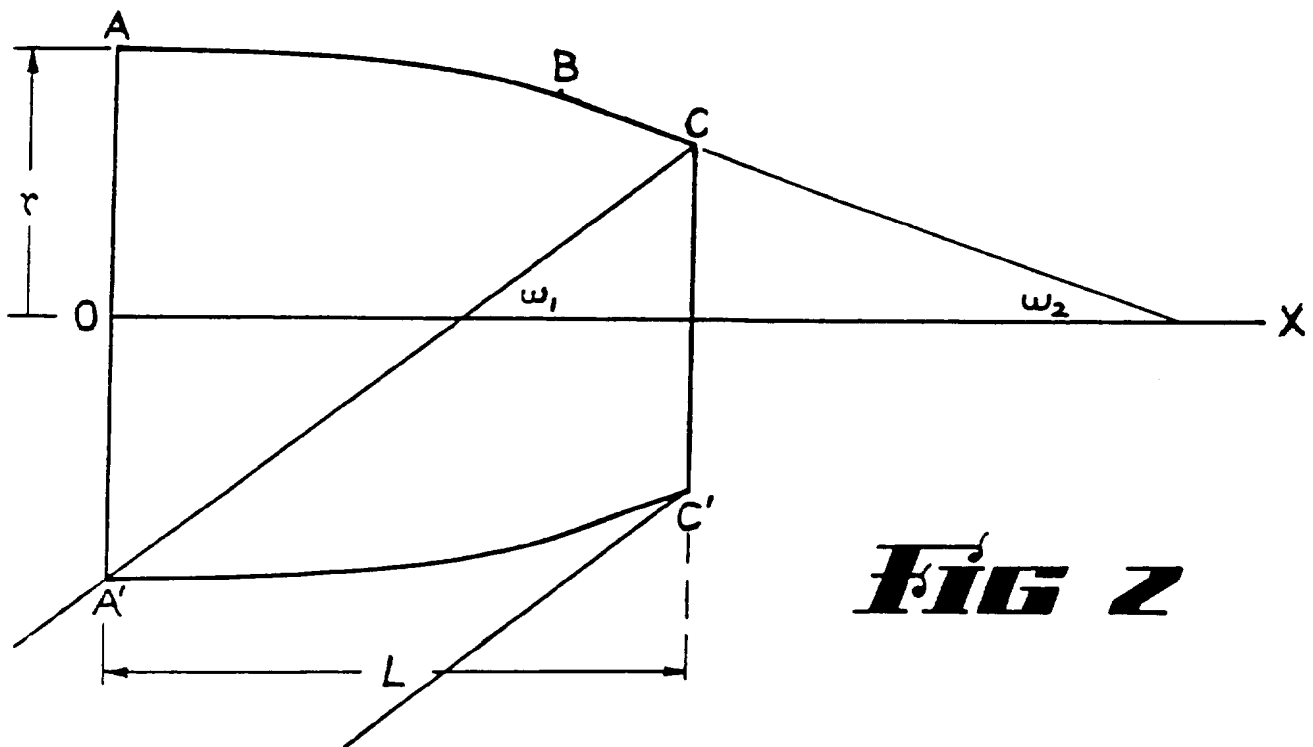





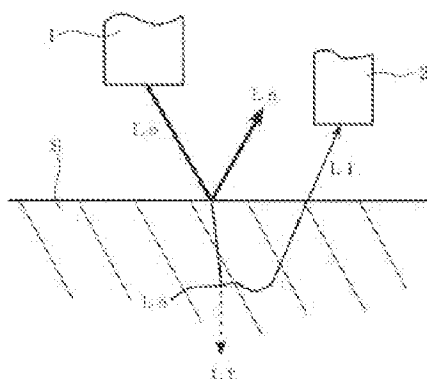
FIG 2

Optical property testing method for skin surface state and device thereof**Publication number:** CN1182572 (A)**Publication date:** 1998-05-27**Inventor(s):** HARUTO KAZAMA [JP]; YOSHINAO NAGASE [JP]; YUKIBAKU YATA [JP]**Applicant(s):** KAO CORP [JP]**Classification:****- international:** **A61B5/00; A61B5/103; A61B5/107; A61B5/1455; A61B10/00; A61B5/00; A61B5/103; A61B5/107; A61B5/145; A61B10/00;** (IPC1-7): A61B5/00; G01N21/00**- European:** A61B5/103N4**Application number:** CN19971022796 19971025**Priority number(s):** JP19960301277 19961025; US19980176026 19981021**Also published as:** CN1161073 (C) US6070092 (A) JP10127585 (A)

Abstract not available for CN 1182572 (A)

Abstract of corresponding document: **US 6070092 (A)**

Light L_o is allowed to be applied on the skin S , the internal scattered light L_i in the region from the surface of the skin S to the microcirculation system is detected preferably in at least three wavelength regions of 400 to nm, 500 to 600 nm, 600 to 800 nm, or 800 to 1500 nm, and the optical properties which are affected by blood circulation, such as skin color, spots, and discoloration are determined. With these optical properties, the surface state of the skin can be evaluated better.

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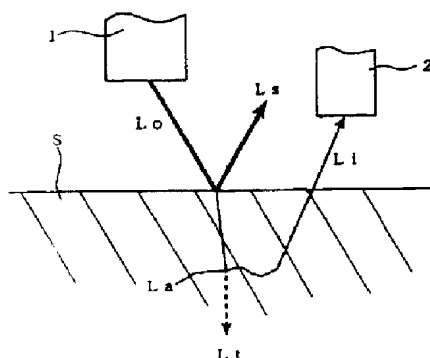
代理人 叶恺东 萧掬昌

权利要求书 2 页 说明书 12 页 附图页数 6 页

[54]发明名称 皮肤表面状态的光学特性测定方法及其装置

[57]摘要

本发明可以用于正确地评价受血液循环影响的皮肤肤色合适程度,病患、光泽、灰暗等光学特性;方案是使光 L_o 入射在皮肤 S 上,检测从皮肤表面到微循环系统范围的内部散射光 L_i ,计算出光学特性,借此测定皮肤表面状态。这时作为检测光最好使用波长为 400~500nm, 500~600nm、600~800nm、800~1500nm 的三个以上波长范围的光。



权 利 要 求 书

1、一种皮肤表面状态的光学特性测定方法，其特征在于：使光入射在皮肤上，通过检测从皮肤表面到微循环系统范围内的内部散射光计算出光学特性。

5 2、如权利要求1所述的测定方法，其特征在于：使光发射单元与皮肤接触发出入射光，使光接收单元接触皮肤接收皮肤的内部散射光。

3、如权利要求1所述的测定方法，其特征在于检测波长是400 ~ 1500nm的光。

4、如权利要求1所述的测定方法，其特征在于检测波长为400 ~ 500nm
10 的光。

5、如权利要求1所述的测定方法，其特征在于检测波长当500 ~ 600nm的光。

6、如权利要求1所述的测定方法，其特征在于检测波长为600 ~ 800nm的光。

15 7、如权利要求1所述的测定方法，其特征在于检测波长为800 ~ 1500nm的光。

8、如权利要求3、4、5、6或7所述的测定方法，其特征在于：通过检测3个以上波长范围的光，计算出与皮肤的血红蛋白含量有关光学的特性。

9、如权利要求3、4、5、6或7所述的测定方法，其特征在于：通过检
20 测3个以上波长范围的光，计算出与血红蛋白氧饱和度有关的光学特性。

10、如权利要求3、4、5、6或7所述的测定方法其特征在于、通过检测3个以上波长范围内的光，计算出与皮肤黑色素含量有关的光学特性。

11、如权利要求3、4、5、6或7所述的测定方法，其特征在于通过检测3个波长范围的光计算出皮肤的肤色指数。

25 12、一种实施权利要求1 ~ 11中任意一项方法的皮肤表面状态的测定装置，其特征在于：包括使确定波长范围的光入射到皮肤上的发光单元，接收从发光单元入射到皮肤上的光的内部散射光的光接收单元，根据由光接收单元接收的光的波长和光量计算出从皮肤表面到微循环系统范围的皮肤光学特性的计算单元；该发光单元和光接收单元上与皮肤相对的那一面被高度为0.2 ~ 0.8mm的周
30 壁包围。

13、如权利要求 12 所述的测定装置，其特征在于：发光单元可以使 3 个以上波长范围的光入射。

说明书

皮肤表面状态的光学特性测定方法及其装置

5 本发明涉及通过计算从皮肤表面到微循环系统受到血红朊、黑色素等影响范围的光学特性测定皮肤表面状态的方法。

现有的测定皮肤色素沉着、色泽灰暗等评价方法。是使特定波长的光入射到皮肤表面上，计算接收到的该反射光特定波长中的皮肤反射率借此评价红斑和色素沉着的方法是公知的（特开昭 58-33153 号公报）。另外，为了测定皮肤的色
10 素沉着而通过计算在特定的两个波长下的皮肤反射率的比值计算出关于黑色素含量和血红朊含量等指数的方法也是公知的（特开平 5-16160 号公报）。

另外，作为不受血液影响的表皮色相测定方法，通常是吸引皮肤表面将部分拉起，使白光射入到该拉起部分然后计算出表皮透射光的分光谱的方法也是公知的（特开平 5-87730 号公报）。

15 可是，利用上述测定皮肤的反射光测定方法，由于在接收光中除了含有皮肤内部的散射光之外还会有许多其它表面散射光，所以在接收光的光路长度上的色散增加，引起不能正确分析的问题。

另外在检测表皮透射光的方法中存在由于排除血液对检测结果的影响而不能评价受血液循环影响的皮肤表面的色度程度的问题。

20 本发明的目的是解决已有技中的问题，正确地评价血液微循环状态和由于黑色素沉着引起的有关色泽灰暗、黑影等皮肤的光学特性。

本发明人发现通过把从皮肤表面到微循环系统的范围作为测定对象，检测皮肤内部散射光可以达到上述目的，从而完成了本发明。

也就是说，本发明提供一种特征为使光入射到皮肤上、检测从皮肤表面到微
25 循环系统范围的内部散射光然后计算光学特性的皮肤表面状态的光学特性测定方法。

特别是用在波长 400 ~ 1500nm 的范围内互相不同的三个以上波长范围的光入射并且检测，或者用分别属于波长 400 ~ 500nm，波长 500 ~ 600nm 和波长 600-800nm 波长范围的三个波长范围的光入射并且检测，借此计算光学特性，
30 从而提供一种皮肤表面状态的测定方法。

另外，作为实施这些方法的装置，本发明提供一种包括使确定波长范围的光入射在皮肤上的光发射单元、接收从光发射单元入射到皮肤并接收内部散射光的光接收单元和根据由光接收单元接收的光的波长和光通量计算从皮肤表面到微循环系统范围的皮肤光学特性的计算单元并且使与皮肤相对的光发射单元和光接收单元的表面被高度为 0.2 ~ 0.8mm 的周壁围住的皮肤表面状态测定装置。

按照本发明，由于检测从皮肤表面到微循环系统范围的内部散射光，所以可以正确地检测到在皮肤的角质层中存在的并且作为皮肤色素沉着原因的黑色素和在微循环系统内的血红朊。从而可以正确评价反应皮肤色泽灰暗和血流循环的皮肤表面色度情况等皮肤表面状态。

另外，按照本发明，由于不是检测表面散射光，而是检测内部散射光，所以检测光的光路长度的偏差较小，从而使 S/N 比提高。

另外，在本发明中，所谓皮肤的微循环系统是毛细血管与组织间的边界范围的血流系统，对于全吸光度（组织的吸光度，毛细血管的静脉血吸光度和伴随脉动的动脉血的吸光度的总合），伴随脉动吸光度成分称为 5% 以下的最小血流系统。

如本发明所述，如果把伴随脉动吸光度成分为 5% 以下的微循环系统作为检测对象，则可以评价受血液循环影响的皮肤表面色度的情况。对此，如果不是把位于对皮肤色泽有影响的非常浅部的血流作为检测对象，而是把伴随脉动吸光度成分（动脉血）作为主要检测对象，则因为只能检测出深部血液的光学特性，所以不能测定受血液循环影响的皮肤颜色。

图 1 是本发明方法测定原理的说明图。

图 2 是氧化型血红朊（ HbO_2 ）和还原型血红朊（ Hb ）的吸收谱图。

图 3 是皮肤色病患的色分光特性图。

图 4 是本发明装置实例的概略构成图[图（a）]及其发射/接收光探测器的底面图（图（b）]

图 5 是与本发明不同装置实例的使用状态的说明图。

图 6 是在实施例中的亮度指数 L^* 与黑色素指数的关系图。

图 7 是比较例中的亮度指数 L^* 与黑色素指数的关系图。

图 8 是压住时间与总的血红朊浓度或氧化饱和度的关系图。

图 9 是靠人感官评价仿纹与实施例的皮肤色指数的关系图。

下面结合图面详细说明本发明的方法及其装置，在各图中凡同一符号表示相同的构成部件。

图 1 是本发明方法测定原理的说明图。在该图中当光 L_0 从发光单元 1 入射到皮肤 S 上时，便分成入射光 L_0 表面散射光 LS 和入射在皮肤 S 内部的光，射入皮肤内部的光分成透过真皮方向的光 L_t 、吸收的光 La 和内部散射光从皮肤表面射出的光 Li 。如果入射到皮肤 S 上的光 L_0 是可见光，则由于真皮以下的组织的透射光 L_t 很少，所以透射光 L_t 与表面散射光 LS ，内部散射光 Li 和吸收光 La 的光量相比可以略去不计。因此

[数 1]

可以近似表示为：入射光量=表面散射光光量+内部散射光光量+吸收光光量。

因为随着皮肤内部的色素吸收波长和色素浓度波长分布及光量分别对应变化，所以可以认为内部散射光和吸收光具有根据皮肤内部色素变化的信息。

因此，在上述的公式中可以略去表面散射光或者作为噪声处理，以便可以使测定系统在吸收、入射光光量和内部散射光上适用 Lamber-beer 定律，可以用下式表示

[数 2]

$$\log(I_0/I_s) = \epsilon C l$$

(式中 I_0 : 入射光光量, I_s : 内部散射光光量, ϵ : 色素的吸光系数, C : 色素浓度, l : 光路长度)

可是，作为影响内部散射光的皮肤内部的色素，例如可以举出主要存在于表皮中的黑色素，存在于真皮中的血红朊，其中血红朊包括氧化型 (HbO_2) 和还原型 (Hb)。因此，这些色素具有图 2 所示的固有吸收谱。另外皮肤的细胞组织对内部散射光的影响可以略去不计。

因此，通过检测皮肤内部的散射光可以计算出皮肤内部的血红朊、黑色素等色素的浓度

例如在图 2 中，由于在 565nm(G (绿))和波长 830nm (IR (红外))的波长下，氧化型血红朊与还原型血红朊的吸光系数相等，所以可以简化根据检测光的吸光度求解浓度的计算。另外，根据氧化型血红朊 (HbO_2) 与还原型血红朊 (Hb) 在波长 660nm (R (红))下的吸光系数的比率最大，通过与在上

述氧化型血红蛋白与还原型血红蛋白的吸光系数相等的波长（565nm，830nm）下的吸光度的比较可以计算出血红蛋白的氧饱和度。因此，设波长565nm（G（绿）），830nm（IR（红外）），660nm（R（红））三个波长下的吸光度分别为AG，AIR、AR，考虑对这三个波长吸光度的各个色素的关系可以计算出各个色素的浓度。

在这种情况下，黑色素，血红蛋白分别存在于表皮和真皮中，表皮的厚度为 d_b ，真皮的厚度为 d_e 。

[数3]

设 $d_b = K d_e$ 。（式中K为系数）

另外，由于波长G（绿），IR（红外），R（红）的透过深度不同，所以可以将

[数4]

表示为 $d(R) = m \cdot d(G)$ ， $d(IR) = n \cdot d(G)$

在用下标D表示真皮，下标E表示表皮时这些关系也成立。

因此上述三个吸收光度AG，AIR，AR分别用如下的（1）～（3）式表示。式中： $\epsilon_X(Y)$ 表示在波长为Y时的X吸光系数，GX表示X的浓度。

[数5]

因为 $d_{D(G)} = K \cdot d_{E(G)}$

所以 $A_G = A_{HbO2(G)} + A_{Hb(G)} + A_{M(G)}$

$$\begin{aligned}
 &= \epsilon_{HbO2(G)} \cdot C_{HbO2} \cdot d_{D(G)} + \epsilon_{Hb(G)} \cdot C_{Hb} \cdot d_{D(G)} + \epsilon_{M(G)} \cdot C_M \cdot d_{E(G)} \\
 &= \epsilon_{HbO2(G)} \cdot (C_{HbO2} + C_{Hb}) \cdot K \cdot d_{E(G)} + \epsilon_{M(G)} \cdot C_M \cdot d_{E(G)} \\
 &= (K \cdot \epsilon_{HbO2(G)} \cdot C_{HbO2} + \epsilon_{M(G)} \cdot C_M) \cdot d_{E(G)}
 \end{aligned} \tag{1}$$

[数6]

因为 $d_{E(R)} = n \cdot d_{E(G)}$ ， $d_{D(R)} = K \cdot n \cdot d_{E(G)}$

所以 $A_R = A_{HbO2(R)} + A_{Hb(R)} + A_{M(R)}$

$$\begin{aligned}
 &= \epsilon_{HbO2(R)} \cdot C_{HbO2} \cdot d_{D(R)} + \epsilon_{Hb(R)} \cdot C_{Hb} \cdot d_{D(R)} + \epsilon_{M(R)} \cdot C_M \cdot d_{E(R)} \\
 &= \epsilon_{HbO2(R)} \cdot (C_{HbO2} + C_{Hb}) \cdot K \cdot n \cdot d_{E(G)} + \epsilon_{M(R)} \cdot C_M \cdot n \cdot d_{E(G)} \\
 &= (K \cdot n \cdot \epsilon_{HbO2(R)} \cdot C_{HbO2} + \epsilon_{M(R)} \cdot C_M) \cdot d_{E(G)}
 \end{aligned}$$

(2)

[数 7]

因为 $d_{E(R)} = m \cdot d_{E(G)}$ 、 $d_{D(R)} = K \cdot m \cdot d_{E(G)}$

所以 $A_R = A_{HbO2(R)} + A_{Hb(R)} + A_{M(R)}$

$$\begin{aligned} 5 \quad &= \varepsilon_{HbO2(R)} \cdot C_{HbO2} \cdot d_{D(R)} + \varepsilon_{Hb(R)} \cdot C_{Hb} \cdot d_{D(R)} + \varepsilon_{M(R)} \cdot C_M \cdot d_{E(R)} \\ &= \varepsilon_{HbO2(R)} \cdot C_{HbO2} \cdot K \cdot m \cdot d_{E(G)} + \varepsilon_{Hb(R)} \cdot C_{Hb} \cdot K \cdot m \cdot d_{E(G)} + \varepsilon_{M(R)} \cdot C_M \cdot m \cdot d_{E(G)} \\ &= (K \cdot m \cdot \varepsilon_{HbO2(R)} \cdot C_{HbO2} + K \cdot m \cdot \varepsilon_{Hb(R)} \cdot C_{Hb} + m \cdot \varepsilon_{M(R)} \cdot C_M) \cdot d_{E(G)} \end{aligned}$$

(3)

10 因此, 根据式 (1) ~ (3), 可以将氧化型血红朊浓度 C_{HbO2} , 还原型血红朊浓度 (Hb), 黑色素浓度 C_M 表示成下式 (4) ~ (6)

【数 8】

$$\begin{aligned}
 C_{\text{H}_2\text{O}} \cdot d &= \frac{(\epsilon_{\text{H}_2\text{O}}(\text{IR}) \cdot \epsilon_{\text{M}}(\text{R}) - \epsilon_{\text{H}_2\text{O}}(\text{R}) \cdot \epsilon_{\text{M}}(\text{IR})) A_{\text{G}} + 1/n \cdot (\epsilon_{\text{H}_2\text{O}}(\text{R}) \cdot \epsilon_{\text{M}}(\text{G}) - \epsilon_{\text{H}_2\text{O}}(\text{G}) \cdot \epsilon_{\text{M}}(\text{R})) A_{\text{IR}}}{\epsilon_{\text{H}_2\text{O}}(\text{G}) \cdot \epsilon_{\text{M}}(\text{R}) - \epsilon_{\text{H}_2\text{O}}(\text{R}) \cdot \epsilon_{\text{M}}(\text{G})} + 1/m \cdot A_{\text{R}} \\
 &= \frac{-10500 A_{\text{G}} - 1390000/n \cdot A_{\text{IR}}}{554000} + 1/m \cdot A_{\text{R}} \\
 &= \frac{105 A_{\text{G}} + 13900/n \cdot A_{\text{IR}}}{5540} - 1/m \cdot A_{\text{R}} \\
 &= \frac{-11.8 \cdot K}{11.8 \cdot K}
 \end{aligned}$$

(4)

【数 9】

$$C_{H_2O} \cdot d = \frac{\frac{(\epsilon_{H_2O_2(R)} \cdot \epsilon_{M(R)} - \epsilon_{H_2O_2(R)} \cdot \epsilon_{M(R)}) A_G + 1/n \cdot (\epsilon_{H_2O_2(G)} \cdot \epsilon_{M(R)} - \epsilon_{H_2O_2(R)} \cdot \epsilon_{M(G)}) A_R}{\epsilon_{H_2O_2(G)} \cdot \epsilon_{M(R)} - \epsilon_{H_2O_2(R)} \cdot \epsilon_{M(G)}}}{K \cdot (\epsilon_{H_2O_2(R)} - \epsilon_{H_2O_2(R)})} - 1/m \cdot A_R$$

$$= \frac{\frac{-260000 A_G + 1540000/n \cdot A_R}{554000} - 1/m \cdot A_R}{-11.8 \cdot k} = \frac{\frac{260 A_G - 15400/n \cdot A_R}{5540} - 1/m \cdot A_R}{11.8 \cdot k}$$

(5)

【数 10】

$$C_M \cdot d = \frac{1/n \cdot \epsilon_{HbO_2(G)} \cdot A_{IR} - \epsilon_{HbO_2(IR)} \cdot A_G}{(\epsilon_{HbO_2(G)} \cdot \epsilon_{M(IR)} - \epsilon_{HbO_2(IR)} \cdot \epsilon_{M(G)})} = \frac{195/n \cdot A_{IR} - 3.77A_G}{554000}$$

5

(6)

因此, 通过计算出吸光度 A_G 、 A_R 、 A_{IR} , 可以计算出氧化型血红蛋白浓度 CH_{bO_2} , 还原型血红蛋白浓度 C_{Hb} , 黑色素浓度 C_M 。

按照上述的本发明, 通过计算出血流中的氧化型血红蛋白浓度 CH_{bO_2} , 还原型血红蛋白浓度 C_{Hb} 和氧化型浓度 CH_{bO_2} 与还原型血红蛋白浓度 C_{Hb} 的比率, 可以比较详细地评价血流对皮肤表面颜色的影响。因此氧化型血红蛋白 (HbO_2) 浓度与还原型血红蛋白 (Hb) 浓度的比率是氧饱和度, 即血红蛋白正在与氧结合的比例。另外, 还可以正确评价黑色素沉着的状态。

图 3 是用专门的仪器板通过目视将被检验者 (年龄 20 ~ 50 岁, 30 名) 的皮肤表面分为有色泽灰暗 ($N=17$) 和无色泽灰暗 ($N=14$) 两类, 并且将色泽灰暗有无 (即有无表面颜色病患) 与被检验者的皮肤吸光度的分光特性汇总后的表面病患的色分光特性图。如该图中所示, 根据在波长 500 ~ 600nm(R)附近的吸收峰的大小可以明显地区分出皮肤色病患评价结果。据此认为血液中总的血红蛋白多的人血液流动好, 出现红的健康肤色, 总血红蛋白少的人血液流动不良出现青的肤色。

因此, 按照本发明可以利用作为比较简便的评价指数的“肤色指数”评价与血红蛋白相关连的肤色合适程度。即肤色指数相当图 3 中的斜线面积, 用下式表示。

【数 11】

肤色指数 = $1/2 \times (B \text{ 的吸光度 } (470nm) + G \text{ 的吸光度 } (565nm)) \times (565-470) + 1/2 \times (G \text{ 的吸光度 } (565nm) + R \text{ 的吸光度 } (660nm)) \times 660-565 - 1/2 \times (B \text{ 的吸光度 } (470nm) + R \text{ 的吸光度 } (660nm)) \times (660-470)$
= $95/2 \times (B \text{ 的吸光度 } (470nm) - 2 \times G \text{ 的吸光度 } (565nm) + R \text{ 的吸光度 } (660nm))$ 。

用这个肤色指数评价皮肤色度的合适程度也包含在本发明中。

虽然上面叙述的是利用三个波长接收皮肤内部散射光的评价方法, 但是, 这不是对本发明的限定。检测的波长可以根据光源、光接收传感器等适当设定。例

如可以使用波长为 400 ~ 1500nm 的任意波长, 另外, 也可以使用包含氧化型血红朊和还原型血红朊的吸光系数相等且波长为 400 ~ 500nm 的 B 范围的波长, 氧化型血红朊和还原型血红朊的灵敏度最大为 500 ~ 600nm 的 G 范围的波长, 含有使氧化型血红朊和还原型血红朊的吸光系数相等的波长的 600 ~ 800nm 范围的波长、波长为 800 ~ 1500nm 的 IR 范围的波长中任何一个范围内的波长。

例如在检测波长对三种色素有贡献的情况下, 最好从上述各波长范围中适当选择出 3 个以上波长。

图 4 是通过检测从皮肤表面到微循环系统范围的内部散射光计算出光学特性的本发明装置例子的概略构成图 (图 4 (a)) 及其发射/接收光探测头的底面图 (图 4 (b))。

图 4 中所示的装置包括: 由与皮肤紧密接触发射确定波长的光的光发射单元 1 和接收从光发射单元 1 入射到皮肤 S 上所产生的内部散射光的光接收单元组成一体的发射/接收探测头 3、放大被光接收单元 2 接收的检测信号的放大器 4 和根据检测出的光的波长和光量计算出该皮肤 S 的光学特性的计算机 5。该装置的光发射/光接收探测头 3 的特征为光发射单元 1 上与皮肤相对的一面 1a 和光接收单元 2 上与皮肤 S 相对的一面 2a 被作为发射/接收探测头 3 的外筒延长形成的高度为 0.2 ~ 0.8mm 的周壁所包围。因此, 如果采用这个装置, 则将如下述那样, 在测定时, 即使光发射/光接收探测头 3 的周壁 3a 的边缘 3b 紧密接触在皮肤 S 上, 皮肤 S 的测定部位在光发射单元 1 和光接收单元 2 的压迫下也不会使该部位的微循环系统的血液排开, 从而可以防止在测定时由血红朊引起吸收的降低。与此相反如图 5 所示, 如果光发射单元 1 上与皮肤 S 相对的那一面 1a 和光接收单元 2 上与皮肤 S 相对的那一面 2a 与光发射/光接收探测头 3 的外筒边缘 3b 形成在同一平面上, 则在光发射/光接收探测头 3 与皮肤紧密接触时使测定部位的皮肤 S 受压迫, 使该部位的微循环系统的血液排开, 不可能正确测定皮肤的未状态, 这是不可取的。

在图 4 的装置中, 发光单元 1 由 4 种发光波长的发光二极管 (B (470nm), G (565nm), R (660nm), IR (830nm)) 组成, 这些发光二极管在发光单元 2 的周围配置成同心圆状。为了减少光发射方向的误差, 每种颜色的发光二极管最好设置 2 个以上。

接收单元 2 由光电二极管组成, 虽然在图 4 中的光接收单元 2 是由单一的光

电二极管组成，但是为了提高检测精度，可以根据需要使用若干个光电二极管。

另外，这些光发射单元 1 与光接收单元 2 各自的中心距离 L 通常为 1 - 3mm，以便使测定对象范围能从皮肤表面到微循环系统。

利用这个装置测定皮肤 S 的光学特性时，使光发射/光接收探测头 3 的外筒边缘 3b 与皮肤 S 紧密接触，使在光发射/光接收探测头 3 的内侧形成的与发光单元 1 相对的面 1a 和与光接收单元 2 相对的面 2a 与皮肤 S 接触，这样可通过其一个面 1a 与皮肤接触的发光单元 1 向皮肤 S 发射光，可以防止发射到皮肤上的光在受到皮肤表面凹凸的散射后被光接收单元 2 接收，另外，可以使从光发射单元 1 入射到皮肤 S 内部的入射光量增加，还可以降低由于入射光角度引起的入射光量的偏差。另外，通过使光接收单元 2 的表面 2a 与皮肤 S 接触可以防止皮肤表面的散射光被接收。因此，如果采用这个装置，则因为在从发光单元 1 入射到皮肤 S 的光中，不会接收皮肤表面的散射光，只能接收内部的散射光，所以在根据从皮肤表面到微循环系统范围内的黑色素和血红蛋白等含量检测确定波长的光量时，可以提高 S/N。

另外，如上所述，在测定时使光发射/光接收探测头 3 的边缘 3b 紧密接触在皮肤 S 上的情况下，因为光发射单元 1 上与皮肤 S 相对的那一面 1a 和光接收单元 2 上与皮肤 S 相对的那一面 2a 被光发射/光接收探测头 3 的周壁 3a 包围，周壁 3a 的厚度 3b 为 0.2 - 0.8mm，所以皮肤 S 的测定部位在受到光发射单元 1 和光接收单元 2 的压迫下不会排开该部位的微循环血液，从而可以正确地测定反映受微循环系统影响的皮肤 S 的光学特性。

下面根据实施例具体说明本发明。

比较例 1，实施例 1

使用立即引起皮肤黑化的 UVB（波长 290 - 320nm）照射前腕内侧并分 6 个等级改变照射量进行诱发色素沉着（利用分光型色差计（ミノルタ公司制，CM-1000），获得的亮度指数测色值 L^* =62.383，61.348，61.05，58.853，57.183，56.892），按各照射量等级进行测定，然后求出黑色素指数。

这时在比较例 1 中，利用 MEXAMETER MX-16 Courage & Knazaka 公司制造的接收反射光的测定仪器，用发光二极管 LED 作为光源将波长为 568nm,660nm,880nm 的各波长的光以非接触的方式入射到皮肤上，光接收单元同时检测由发光单元入射到受检体上的光的表面反射光和内部散射光。

另外，作为实施例 1 的测定仪器，采用图 4 中所示的装置，例如在这种情况下，设从光发射单元 1 发射的光是波长为 G（660nm）和 R（830nm）的光。

另外黑色素指数根据用 MX-16 的黑色素计算方法按下式计算出。

[数 12]

5 黑色素指数 = $\text{Log } 5 \times [\text{Log}(R/G) + \text{Log } 5] \times 500$

（式中：R 和 G 分别为接收的光波长 G（660nm）和 R（830nm）的光强度。）

这些结果示在图 6（实施例 1）和图 7（比较例 1）中。

无论在图 6 还是在图 7 中，虽然 UVB 的照射量越大，色素沉着越多，利用
10 分光型测色仪的亮度测色值 L^* 变小，但是从图中可看出，图 6 中所示用实施例 1
检测内部散射光的结果是相关系数 R 明显变大，一次回归式的斜率变大。因此，
可以看出，检测内部散射光的实施例 1 的测定比检测表面反射光的比较例 1 的测定
具有更高的精度和灵敏度。

实施例 2

15 研究在最大血压（220mmHg）以上压迫上腕时的压迫时间与血液状态的变化关系。

在这种情况下，把 560nm, 660nm, 830nm 的 3 个波长作为检测波长，根据
式（4）和（5）计算出氧化血红蛋白浓度和还原血红蛋白浓度，再求出总的血红蛋白
浓度和血红蛋白的氧饱和度。

20 关于光达到的深度系数，在本例中取 $m=1.7, n=3.2$ （这些值随着部位、个人的
不同而变化，在本例中不是对这个值的限定）。另外，用分光型色差计（ミノ
ルタ社制 CM-1000）测定亮度 L^* 。这个结果示在图 8 和表 1 中。在图 8 中，横
轴表示压迫时间纵轴表示总的血红蛋白浓度和血红蛋白的氧饱和度，它们分别表示压
迫开始时间选为 1 时的相对值。

25 表 1

	压迫时间（分）	亮度 L^*
	0	61.9
	1	61.4
	2	61.0
30	3	60.7

从图 8 中可以看出,与总的血红蛋白浓度随时间变化较少的情况相反,氧饱和浓度随时间降低很多。这是由于当动脉和静脉两个血流停止,在血流量不变的状态下消耗氧的结果。这时可以观察到肤色亮度也明显降低。因此,显然,为详细评价在皮肤微循环下的血液状态对肤色的贡献,不但要测量总的血红蛋白的浓度,

5 还要求出氧饱和度。

在检测波长为 2 个波长的情况下,虽然通过利用氧化型血红蛋白与还原型血红蛋白的吸光系数相等的血红蛋白的吸收点可以求出总的血红蛋白浓度,但是氧饱和浓度在理论上不能参照等吸收点以外的波长吸收光度计算出。为此,在求氧饱和度的情况下,象上述实施例那样应至少把 3 个以上波长作为检测波长。

10 实施例 3

根据 B (400 ~ 500nm), G (500 ~ 600nm), R (600 ~ 800nm) 三种波长的内部散射光的吸光度求出作为被检验者为 20 岁-70 岁的女性 103 名的面颊肤色指数。另外通过 9 名专业评审员对这些受试者的肤色状态进行 1 (不良) ~ 9 (良好) 9 个等级的人为感官检查,将感官检查结果与肤色指数的关系

15 绘成曲线,其结果示在图 9 中。

从图 9 中可以看出,本发明的肤色指数与由专业译审人员的感管评价结良好的相符。

按照本发明,可以对受血液循环影响的皮肤肤色适合程度,斑点和光泽暗淡等光学特性进行良好的评价。

说明书附图

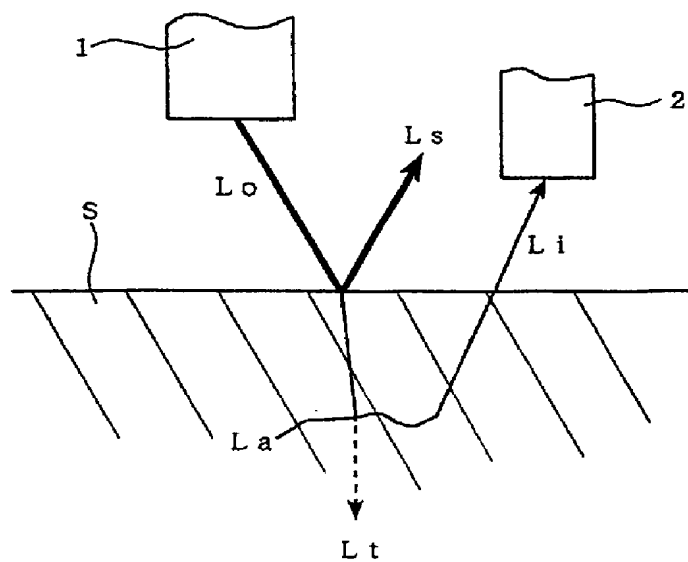


图 1

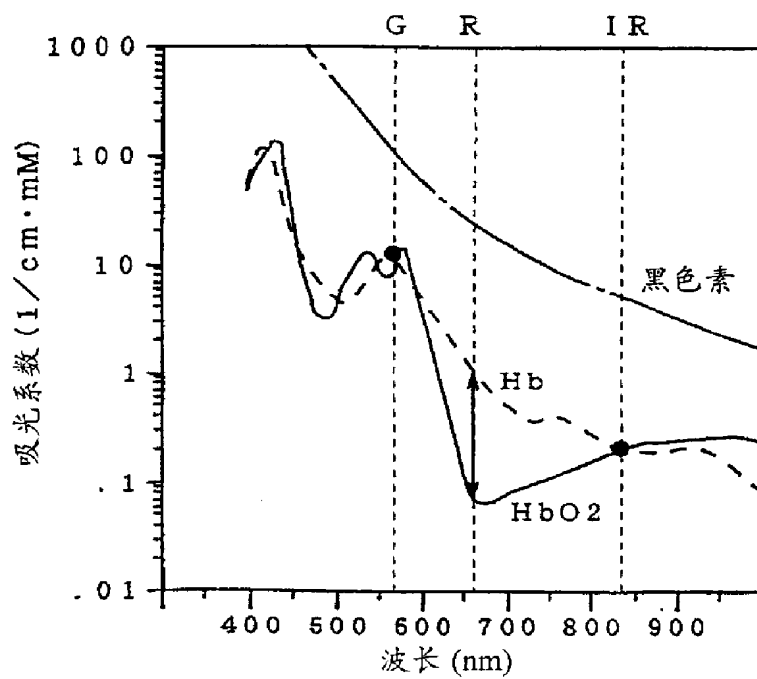


图 2

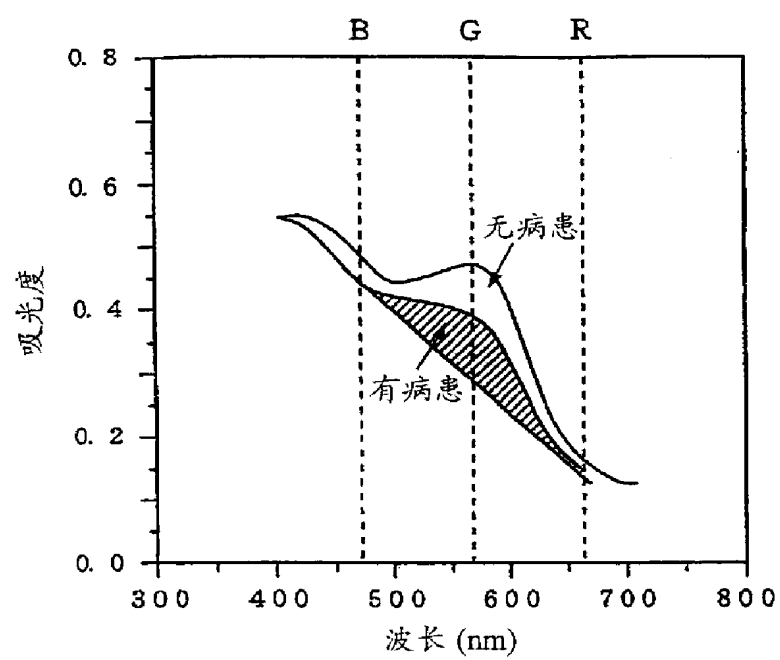
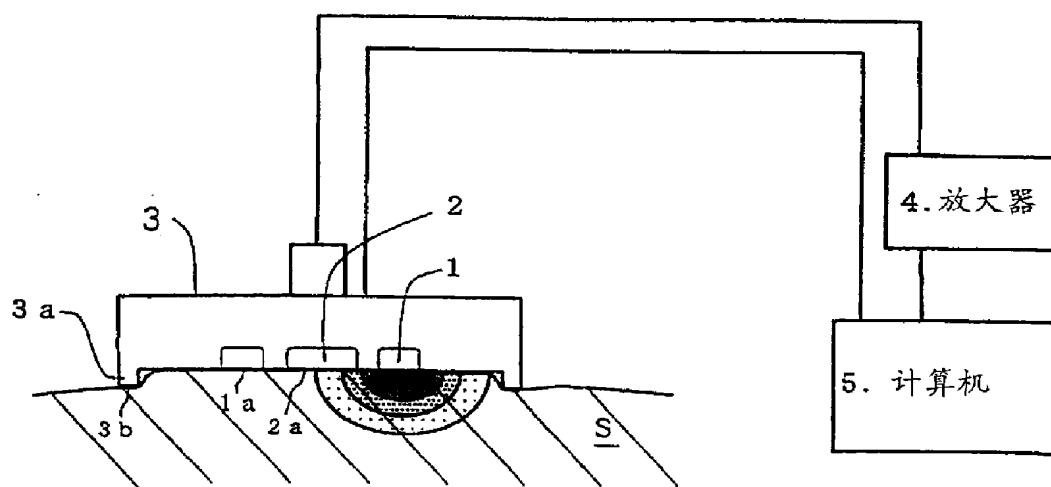


图 3

(a)



(b)

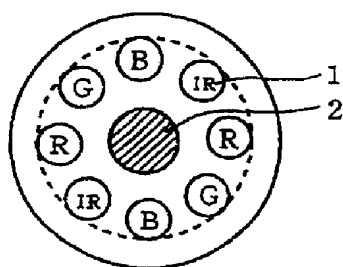


图 4

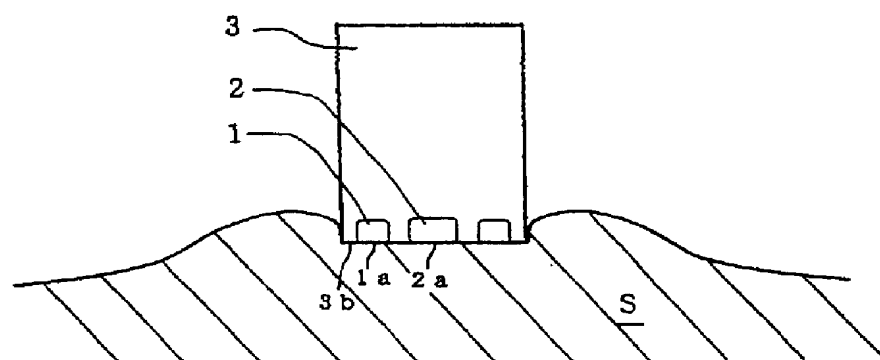


图 5

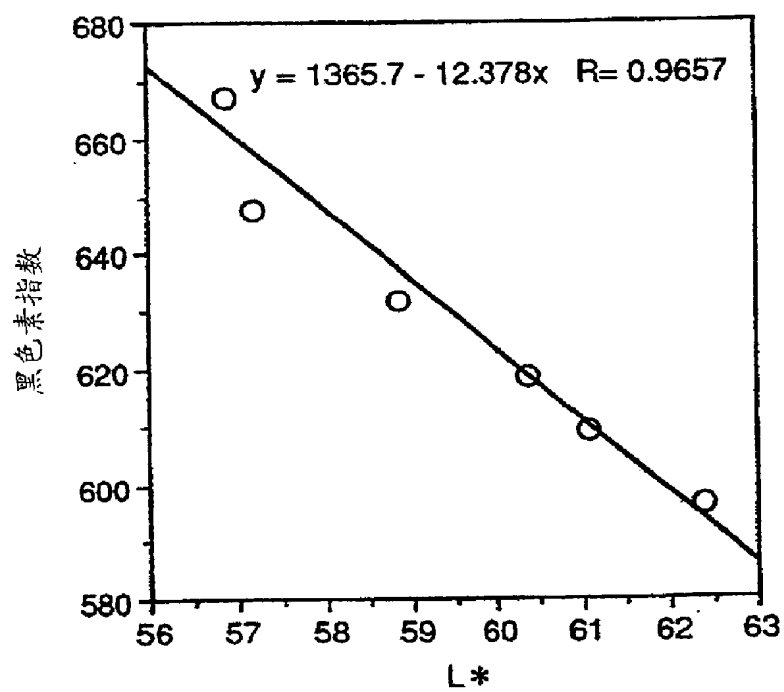


图 6

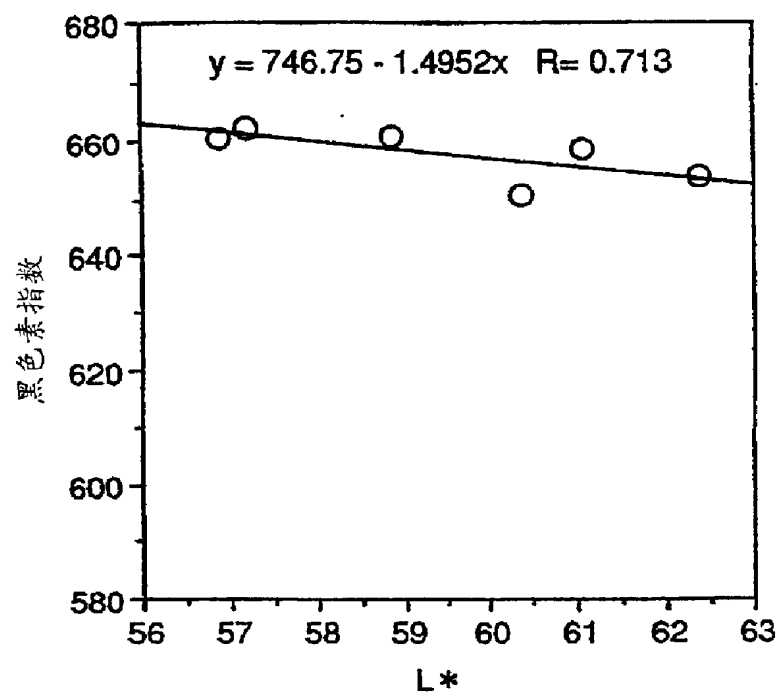


图 7

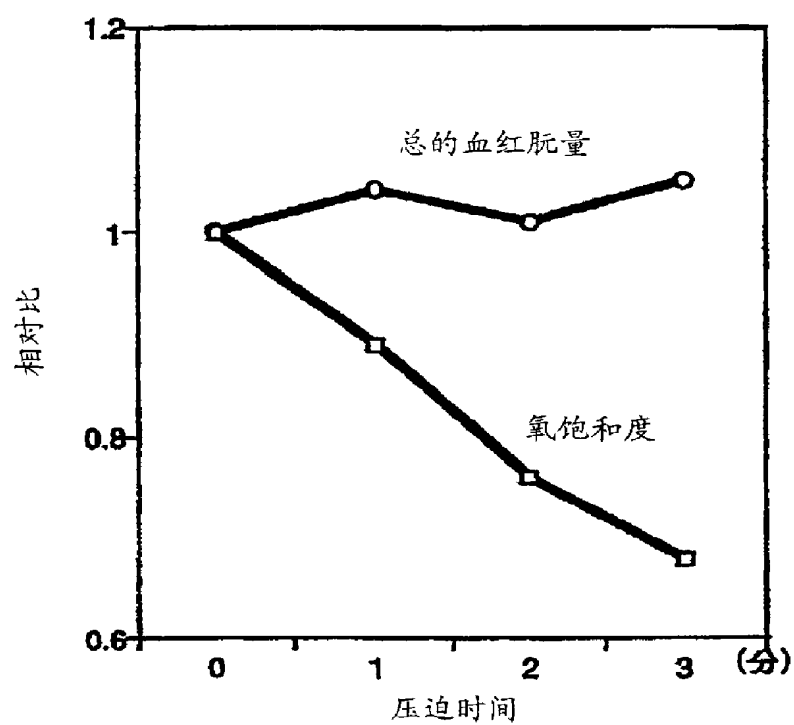


图 8

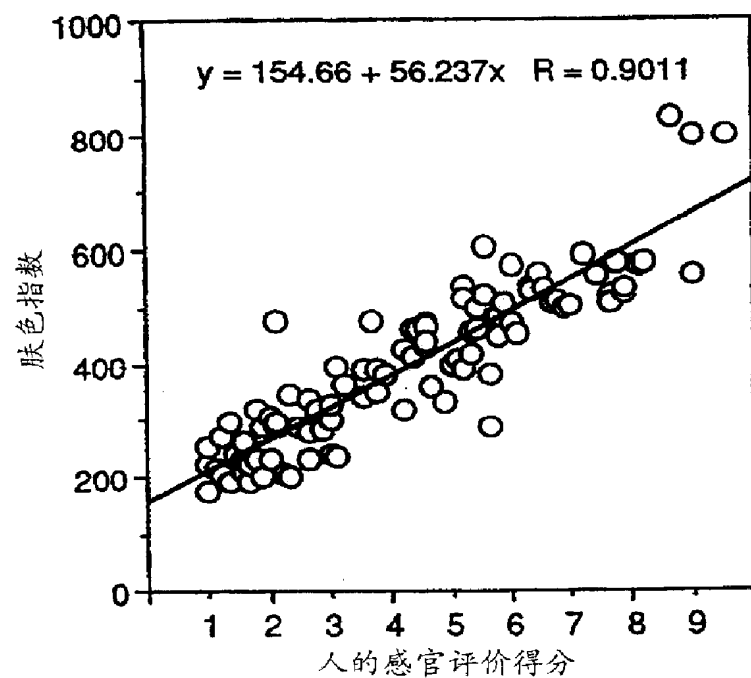







图 9

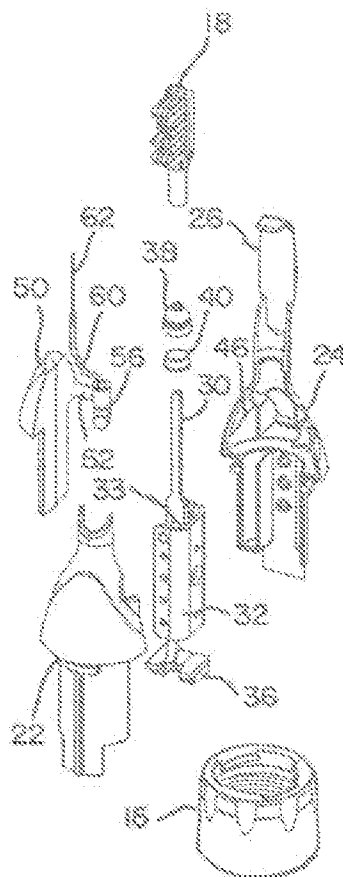
Fluid-dispensing and refilling system for a power toothbrush**Publication number:** CN1351483 (A)**Publication date:** 2002-05-29**Inventor(s):** HALL SCOTT E [US]; LARIMER JAMES N [US]; MILLER KEVIN [US]**Applicant(s):** KONINKL PHILIPS ELECTRONICS NV [US]**Classification:**- **international:** **A61C17/22; A61C17/36; A61C17/16;** (IPC1-7): A61C17/36- **European:** A61C17/22H2; A61C17/36**Application number:** CN20008004977 20000107**Priority number(s):** US19990229979 19990113; US19990404972 19990924**Also published as:** CN1226969 (C) WO0041645 (A1) JP2002534209 (T) EP1143876 (A1) EP1143876 (B1)

more >>

Abstract not available for CN 1351483 (A)

Abstract of corresponding document: **WO 0041645 (A1)**

The fluid delivery/refilling system includes a unit-of-use fluid reservoir for dentifrice or medication within a head portion of a power toothbrush, which is removable from the remainder thereof. A pump element is located in the brushhead and is configured so that the back and forth movement of the brushhead in operation results in fluid moving from the reservoir to a dispensing valve located in the brushhead. The dispensing valve has an end portion which is normally closed, opening under pressure of fluid from the pump. The refilling assembly is separated from the toothbrush and is configured to fit over the top of the toothbrush. A pump used in a power toothbrush for fluid delivery to the bristles. The pump includes a pump chamber with a piston ball located therein. The pump includes a fluid exit channel which connects the discharge end of the pump chamber with an opening in the brushhead in which the dispensing valve is located.

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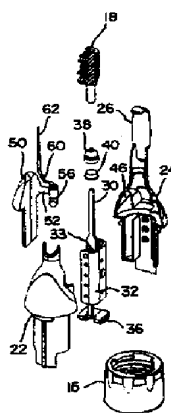
代理人 周备麟 黄力行

权利要求书 5 页 说明书 11 页 附图页数 9 页

[54] 发明名称 用于电动牙刷的流体分配和再充系统

[57] 摘要

本发明的流体输送系统包括一个设置在电动牙刷的头部内的一个使用单位的流体储存器,其可从牙刷的其余部分拆下。一个泵元件设置在刷头中并被如此构造以便工作中刷头的往复运动导致流体从储存器移动到位于刷头中的一个分配阀中。所述分配阀具有一个通常关闭的端部,其在来自泵的流体的压力下打开。再充装置与牙刷分开并被如此构造以便安装在牙刷的顶部。用在电动牙刷中的一个泵用于将流体输送到硬毛。该泵包括一个泵室,一个活塞球设置在所述泵室中。所述泵包括一个流体排出通道,该通道连接泵室的排出端和刷头中的一个孔,所述分配阀设置在刷头中。



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权 利 要 求 书

1. 用于电动牙刷的一种流体输送系统, 包括:

设置在电动牙刷的刷头部分内的储存器装置, 所述头部可从牙刷的把手部分拆下;

5 一个泵元件及相关的流体管路, 用以将流体从所述储存器装置移到牙刷头部的刷头元件;

一个设置在刷头中的分配元件, 该元件连接到所述的流体管路, 在所述泵元件提供的压力下, 容许流体排出到硬毛附近。

10 2. 如权利要求1所述的系统, 包括设置到所述储存器装置外部的一个再充阀元件, 其特征在于: 所述流体管路连接储存器装置、再充阀、泵元件及分配元件。

3. 如权利要求1所述的系统, 其特征在于: 所述泵元件被设置在所述刷头元件中。

15 4. 如权利要求3所述的系统, 其特征在于: 所述泵元件包括一个室和一个球, 在牙刷工作期间该球随刷头的运动在所述室中运动, 其中所述球的运动导致流体从储存器装置进入所述泵元件, 然后从所述泵元件进入所述刷头中的分配元件。

5. 如权利要求4所述的系统, 其特征在于: 所述室是圆柱状的并沿刷头的横向延伸, 其中所述刷头的运动是从一侧到另一侧。

20 6. 如权利要求1所述的系统, 其特征在于: 所述分配元件是一个阀, 其设置在刷头中, 使所述阀的一个自由端部延伸到位于刷头底面之上处, 在刷头硬毛部分的中部。

25 7. 如权利要求6所述的系统, 其特征在于: 所述阀的自由端部包括一对弹性的、通常关闭的唇部, 当流体压力通过泵元件的作用施加于其上时, 它们打开以容许流体从其中排出。

8. 如权利要求1所述的系统, 其特征在于: 所述泵元件设置在一个基座元件上, 所述基座元件在其中包括一个通道, 该通道将泵元件的出口连接到所述分配元件。

30 9. 如权利要求1所述的系统, 其特征在于: 所述储存器装置具有约一个使用单位的流体容量。

10. 如权利要求1所述的系统, 其特征在于: 所述流体是一种牙膏。

11. 如权利要求1所述的系统，其特征在于：所述流体是一种用于口腔医疗的选定药物。

12. 如权利要求2所述的系统，其特征在于：所述再充阀元件包括用于接收来自一个再充装置的针以及一个连接到所述储存器装置的通道的装置，其中流体从外部流体源通过所述针进入所述储存器装置。

13. 如权利要求12所述的系统，其特征在于：所述再充阀包括一个密封部分，该部分防止流体从所述针通过再充阀倒流。

14. 用于电动牙刷的一种流体输送系统，包括：
10 设置在一个牙刷中的储存器装置，所述储存器具有约一个使用单位的流体容量；

一个泵元件及相关的流体管路，用以将流体从所述储存器装置移到牙刷的刷头元件；

15 一个设置在刷头中的分配元件，该元件连接到所述流体管路，在泵装置提供的压力下，容许流体排出到硬毛附近。

15. 如权利要求14所述的系统，其特征在于：所述泵元件设置在刷头中。

16. 如权利要求14所述的系统，其特征在于：所述泵元件包括一个室和一个球，在牙刷工作期间该球随刷头的运动在所述室中运动，其中所述球的运动导致流体从储存器装置进入所述泵元件，然后从所述泵元件进入所述刷头中的分配元件。

17. 如权利要求14所述的系统，其特征在于：所述分配元件是一个阀，其设置在刷头中使所述阀的一个自由端部延伸到位置位于刷头底面之上处，在刷头硬毛部分的中部。

25 18. 一种用在电动牙刷的流体输送系统中的泵，包括：

一个泵元件，用以将流体从一个流体储存器移到位于牙刷刷头部分中的一个分配元件，其中所述泵元件包括一个室和一个球，在牙刷工作期间该球随刷头的运动在所述室中运动，其特征在于所述球在所述室中的运动导致流体从储存器移入泵元件，然后用足够的压力移出泵元件以将流体移到分配元件并在然后从其移出。

19. 如权利要求18所述的系统，其特征在于：所述室是圆柱状的并沿刷头的横向延伸，其中所述刷头的运动是从一侧到另一侧。

20. 一种用于电动牙刷的再充系统，用于和具有一个内置储存器的电动牙刷一起使用，工作期间流体从该储存器移到刷头中的硬毛，内部储存器具有至少一个使用单位的小容积，所述再充系统包括：

一个与电动牙刷分开的外部再充装置，该再充装置具有一个流体
5 储存器，其容纳大量体积的流体，至少是若干使用单位容积的流体；
和再充装置及电动牙刷相联的装置，用于建立一条从再充装置到电动牙刷的流体通道；

用于通过流体通道将流体从再充装置移到电动牙刷的内部储存器中的装置。

10 21. 如权利要求20所述的系统，包括一个位于电动牙刷中的再充阀元件，所述阀元件包括通向所述牙刷中储存器的第一连接件和通向牙刷中一个泵元件的第二连接件。

22. 如权利要求20所述的系统，其特征在于：所述再充装置的结构容许牙刷以一种相互配合的关系插入牙刷中，以便当牙刷以一种选
15 定关系处于再充装置内时，所述流体通道从再充装置建立到电动牙刷。

23. 如权利要求21所述的系统，其特征在于：所述再充装置包括一个再充针，该针在其一端具有一个孔，用于将流体通过再充阀排出到电动牙刷中的内部储存器；在其另一端具有一个孔，用于从再充装
20 置中的储存器接收流体，其中所述牙刷适于安装到再充装置中，以便当再充阀和牙刷处于一种特定的位置关系时，所述针穿入所述再充阀。

24. 如权利要求20所述的系统，其特征在于：所述再充系统包括一个可在两个位置之间移动的顶部元件，其中在一个位置中顶部元件
25 中的一个孔与再充系统其余部分的顶面中的一个孔近似对准，导向其中的储存器，因此再充系统中的储存器可以被方便地再充，而在另一位置，所述顶部元件中的孔完全不与再充系统其余部分的顶面中的孔对准。

25. 如权利要求21所述的系统，其特征在于：所述再充装置包括
30 一个可在其中运动的芯元件，所述芯元件的结构可以容纳包括刷头的牙刷上部，因此当压力被施加到抵靠芯元件的牙刷上时，芯元件运动，首先导致再充装置中的一个中空针元件穿入牙刷中的再充阀，而

且其中抵靠芯元件的刷头的进一步运动导致所述针在再充装置中的进一步运动，所述针具有一个底部和一个位于再充装置的泵室部分中的密封元件，其中当牙刷运动到其在再充装置中的最高位置然后再向下运动时，流体进入泵室，然后进入中空针中，所述流体经过所述针、再充阀并进入牙刷储存器。

26. 一种用在电动牙刷的流体输送系统中的泵，包括：

一个泵元件，用以将提供到该处的流体移送到一个分配元件，所述泵元件适于被设置在牙刷的刷头部分中，其中所述泵元件包括一个泵孔室和一个活塞元件，该元件在牙刷工作期间随刷头的运动在所述室中运动，其中进入泵孔室的流体在活塞元件的作用下沿其运动并从排出端排出，所述泵元件还包括一个刷头中的流体排出通道部分，其在排出端处离开所述泵孔延伸，所述排出通道部分包括连接到所述分配元件的一个位于刷头部分中的孔。

27. 如权利要求26所述的产品，其特征在于：所述泵孔室在其排出端包括一个中凹表面，该中凹表面具有和活塞元件的表面大约相同的曲率，所述活塞元件是球形的，其中工作中的活塞元件在一个泵循环的排出部分的末尾与所述中凹表面接触。

28. 如权利要求27所述的产品，其特征在于：所述中凹表面是一个塞子的一部分，所述塞子形成泵孔的排出端。

29. 如权利要求26所述的产品，包括两个突出部，它们从流体排出通道的纵向相对侧向其内侧延伸，导致流体排出通道的宽度变窄，它们设置在泵孔的排出端和刷头部分中的所述孔之间。

30. 如权利要求29所述的产品，其特征在于：所述突出部被制成锥形以便流体排出通道的宽度沿突出部的整个长度逐渐减小。

31. 如权利要求30所述的产品，其特征在于：所述流体排出通道的最大宽度约为0.10英寸，而且其中所述通道在相隔开的突出部之间变窄到约0.04英寸。

32. 如权利要求26所述的产品，其特征在于：所述泵形成在刷头内部并有一个相配的盖元件。

33. 如权利要求26所述的产品，其特征在于：所述活塞元件由碳化钨制成。

34. 一种用在口腔护理装置的流体输送系统中的泵，包括：

一个泵元件，用以将提供到该处的流体移送到一个分配元件，所述泵元件适于被设置在口腔护理装置的一个移动端部，其中所述泵元件包括一个泵孔室和一个活塞元件，该元件在口腔护理装置工作期间与所述端部的运动响应在所述室中运动，其中进入泵孔室中的流体在活塞元件的作用下沿其运动并从排出端排出，所述泵元件在所述端部还包括一个流体排出通道部分，其在排出端处离开所述泵孔延伸，所述排出通道部分包括连接到所述分配元件的一个位于所述端部中的孔。

35. 如权利要求34所述的产品，其特征在于：所述泵孔室在其排出端包括一个中凹表面，该中凹表面具有和活塞元件的表面大约相同的曲率，所述活塞元件是球形的，其中工作中的活塞元件在一个泵循环的排出部分的末尾与所述中凹表面接触。

36. 如权利要求35所述的产品，其特征在于：所述中凹表面是一个塞子的一部分，所述塞子形成泵孔的排出端。

37. 如权利要求34所述的产品，包括两个突出部，它们从流体排出通道的纵向相对侧向其内侧延伸，导致流体排出通道的宽度变窄，它们设置在泵孔的排出端和所述端部的所述孔之间。

38. 如权利要求37所述的产品，其特征在于：所述突出部被制成锥形以便流体排出通道的宽度沿突出部的整个长度逐渐减小。

39. 如权利要求38所述的产品，其特征在于：所述流体排出通道的最大宽度约为0.10英寸，而且其中所述通道在相隔开的突出部之间变窄到约0.04英寸。

40. 如权利要求34所述的产品，其特征在于：所述泵形成在所述端部内部并有一个相配的盖元件。

41. 如权利要求34所述的产品，其特征在于：所述活塞元件由碳化钨制成。

说明书

用于电动牙刷的流体分配和再充系统

技术领域

5 本发明通常涉及流体分配电动牙刷，尤其是涉及这样一种牙刷，其具有一个体内储存器和再充装置及一个流体通道装置，该流体通道装置包括一个位于牙刷刷头部分中的泵元件。所述泵元件利用移动刷头的动作将流体从体内储存器移到刷头上的硬毛处。

背景技术

10 能防止或杀死引起牙科疾病的细菌的口部牙膏及/或药物的有效输送一直是需要的而且已经成为大量研究努力的目标。已经研制了多种装置用以完成牙膏或药物到硬毛的输送，所述装置包括主动的和被动的。这些装置即包括手动的又包括电动的牙刷。手动牙刷通常使用一个手操作泵，例如Boscardin等的美国专利US4,221,492和Gingras
15 等的美国专利US4,413,370，而电动牙刷使用多种手段，包括体内及体外的主动泵装置，它们包括各种机械的、气动的及液压元件。

一些电动牙刷流体分配系统利用牙刷自身的动作将牙膏从储存器抽到硬毛处，例如Giuliani等的美国专利US5,309,590中所示。一些用于手动和电动牙刷的流体分配系统的代表例包括Balamuth的美国
20 专利US3,547,110；English等的美国专利US5,066,155；Lusting的美国专利US5,208,933及Kuo的美国专利US5,062,728。

用于牙膏的储存器通常设置在牙刷的把手部分，例如'590专利中所示，或者设置在一个单独的装置中，例如'110专利中所示。在一些
25 情况中，储存器是一次性的，因此当储存器中的流体被用完时，储存器本身被抛弃并插入一个新的储存器，这样一种系统见'370专利。在其余的系统中，储存器是可再充的。通常，储存器包含用于流体牙膏或药物的大量特殊应用的足够流体。

用于将流体从刷头进行分配的各种系统也是已知的。在一些情况
30 中，流体从刷头上的许多孔分配，例如'492专利中所示，而在其余的情况中，流体通过硬毛本身分配，例如Evans的美国专利US4,039,261中所示。

但是，流体分配系统往往难以操作，而且是不可靠和昂贵的。许多这种系统太复杂了以致不能可靠操作，而且/或不能正常工作，而其余系统由于流体管路或排出孔的堵塞或差的泵送作用等原因而不能工作。另外，难以设计一种系统，该系统具有一个足够小的可以装
5 在一个普通牙刷结构中的泵。一般使用外部泵，它们体积大而且昂贵。这些泵元件，无论是体内的还是体外的，通常都太复杂了以致不能持续可靠地工作，或者太昂贵而不实用了。鉴于所有这些原因，用于手动和电动牙刷的流体分配系统还没有特别成功。

由于潜在的利益，需要在电动牙刷中设置一个流体分配系统，该系统可靠而且设计足够简单以使制造和维修是切实可行的。

本发明的公开内容

因此，本发明包括用于电动牙刷的一种流体输送系统，其包括：一个设置在电动牙刷的牙刷头部内的储存器，其中所述头部可从牙刷的把手部分拆下；一个泵元件及相关的流体管路，用以将流体从储存
15 器移到牙刷头部的刷头部分；以及一个设置在刷头中的分配元件，该元件连接到所述的流体管路，在所述泵元件提供的压力下，用以从泵元件接收流体并容许流体从其排出到硬毛附近。

本发明还包括用在电动牙刷的流体输送系统中的一种泵，其包括：一个泵元件，用以将流体从一个储存器移到位于牙刷刷头部分中的一个分配元件，其中所述泵元件包括一个室和一个球，在牙刷操作
20 期间该球与刷头的运动响应在所述室中运动，其中所述球在室中的运动导致流体从储存器移入泵元件，然后用足够的压力移出泵元件以将流体移到分配元件并在然后从其移出。

本发明还包括一种用于电动牙刷的再充系统，用于和具有一个内置储存器的电动牙刷一起使用，工作期间流体从该储存器移到刷头中的硬毛，内部储存器具有至少一个使用单位的小容积，所述再充系统
25 包括：一个与电动牙刷分开的外部再充装置，该再充装置具有一个流体储存器，其容纳大量体积的流体，至少是若干使用单位容积的流体；与再充装置和电动牙刷相关的装置，用于产生一个从再充装置到电动牙刷的流体通道；以及用于将流体从再充装置移到电动牙刷的内部储存器中的装置。

附图的简要描述

图1是一个牙刷的示意图，该牙刷包含本发明的流体分配系统。

图2是图1牙刷头部的一个局部分解图。

图3是一个示意图，表示本发明的流体分配系统的一部分。

图4A和4B是本发明的流体分配系统的一个再充阀的横剖图，其中
5 具有和没有一个再充针（needle）。

图5是一个局部分解图，表示泵元件和刷头。

图6和7是表示流体在储存器和刷头之间运动的横剖图，包括移动
流体的泵元件。

图8是表示流体输送系统分配阀的横剖图。

10 图9到13表示本发明的再充装置及再充图1所示牙刷的体内储存
器的过程。

图14是再充装置的一个顶视图。

图15是图5至8所示泵元件的另一个实施例的局部分解图。

图16是图15所示泵元件的一个横剖图。

15 图17是图15所示泵元件的另一个横剖图。

图18是图15所示泵元件的泵室部分的一个横剖图。

图19是图15所示泵元件顶部的一个示意图。

图20是图15所示泵元件的泵室部分的另一个横剖图。

实施本发明的最佳方式

20 图1表示一个典型的电动牙刷，其中使用了本发明的流体分配和
再充系统。但是，应该强调的是本发明并不限于所示和所述的特定电
动牙刷或者限于一种特定的刷头运动。而且，本发明的原理可以应用
于多种结构的电动牙刷，本发明的某些方面甚至可以用在手动牙刷
中。

25 以标号10表示的图1所示的电动牙刷包括一个把手部分12和一个
可动头部14。在把手部分中的是一个电动单元，包括驱动刷头18的一
个电磁驱动器和一个电池，刷头以一种左右方式安装在一个可转动的
臂上，这在下面将要详细论述。牙刷10由一个按钮打开/关闭开关19
控制。

30 图2是一个分解图，表示图1牙刷头部14的基本元件。所述头部包
括两个配合的主体元件22和24，其围绕一个支承元件32装配在一起。
主体元件24包括一个用于刷头18的伸出的护罩26。刷头18安装在一个

转臂30上,接着该臂被安装在一个扭转销轴元件33周围以便转动,元件33在支承元件32的相对侧之间延伸.转臂30由牙刷把手部分12中的一个电磁驱动器(未示出)驱动,该驱动器作用在转臂30端部的磁元件36上.有关这样一种牙刷结构和操作的更详细描述见Giuliani等的美国专利US5,378,153,其具有和本发明相同的受让人。

橡胶密封38和塑料环40将臂30密封到主体元件22和24,防止潮气返回头部14.一个螺母部分15通过螺纹连接将头部14连接到把手部分12.位于一个内腔46中的是一个体内储存器50,该内腔由两个配合的主体部分22和24形成.体内储存器50一般是箭头状的,大约两英寸长,具有沿其周边连在一起的前后类似的弹性塑料板.该实施例所示的体内储存器50通常容纳一次刷牙所需的足够流体,即一个“使用单位”.这种较小的体内储存器容积是重要的,原因在于它容许一个流体储存器设置在图1所示牙刷结构的可动头部14中.这种设置具有许多优点,包括一个较短的流体通道和一个整体简化的结构.因此,当替换头部14时,例如当刷头18的硬毛磨损时,储存器50及相关的流体输送系统是替换单元的一部分。

参照图2和3,从储存器50延伸的是一个短管部件52,该部件装配到一个再充阀56的下端,这在下面详细描述.位于管部件52入口上面、从再充阀56侧面延伸的是一个弹性流体管路60,在该实施例中的是一个直径约0.10英寸、长约1英寸的弹性塑料材料的管路.流体管路60连接到一个流体入口管62,管62的直径约为0.05英寸并延伸到其邻近端64处的刷头18中。

图5至8表示刷头18中的流体输送结构,包括一个流体泵66、一个位于基座元件76中从泵66延伸的流体输送腔68、及一个流体分配阀70,该阀从输送腔68向上延伸,经过刷头表面71上的刷头18进入牙刷的硬毛区域72。

流体入口管62延伸到刷头18中,在该处它连接流体泵66的入口区域83(图7).流体泵66是一个不规则的块,具有一个从其横向延伸的柱状孔75.流体泵66设置在细长的基座元件76的一端.细长的基座元件76约0.03英寸厚并具有一个挖去部分或腔部68,该部68从泵66大约延伸到基座元件76的另一端.该实施例中所示的腔部68约为0.015英寸深。

如图5所示，该实施例中所示的流体泵66和基座元件76形成一个整体元件，该元件装入刷头下表面中的一个相匹配的腔77中。腔77被如此构造以便当该整体元件正确地设置在腔77中时提供一个用于泵66中柱状孔75端部的固体密封。流体输送腔68连接泵66的出口部分85和分配阀70的下端。当一体的泵和基座元件就位时，一个围绕基座元件76中腔68的唇部84接触刷头中的一个配合表面87，提供一个围绕腔68的液密密封。这样腔68就作为一个连接泵66和分配阀70的流体输送装置。

泵66被构造的具有可以被解除的相对拐角，被解除的部分分别从柱状孔75的相对开口端绕过该块体的相邻拐角延伸到一个位置，该位置近似位于沿泵66的邻近侧的中央。这种结构从图6和7中看得最清楚。用作活塞的一个圆球63安放在气缸孔75中。入口区域83重叠孔75的一段，而出口区域85重叠孔75的另一端。出口区域85连接到基座元件中的流体输送腔68。因此，存在一个从管62到腔68的完整流体通道。球63在柱状孔75中移动，当工作中刷头从左到右运动时，对进入孔75的流体产生一种泵送作用。

对于具有作用到阀70的足够流体压力的正确的泵送操作，在球63和柱状孔75的内表面之间留有适当的间隙是重要的。在所示实施例中，该间隙约为0.0015英寸。

操作中，在排出冲程期间，当球向孔75的排出或出口区域85运动时，流体即朝出口区域运动又绕过活塞朝孔75的入口区域83运动。在活塞的返回冲程（朝向入口区域83），关闭的分配阀强迫流体绕经球朝向出口区域，准备下一个排出冲程。流率不完全取决于刷头运动的大小，且因此是牙刷载荷的函数。当刷头系统是一个共振系统时，例如'153专利中所述的，当硬毛与牙齿接触时，流率增加。在流体通道的端部有或没有分配阀都会出现泵送作用，而且工作中的泵送作用是自起动的，即它能泵送空气。

由活塞头的往复运动引起的活塞球63的往复动作迫使流体从泵的出口区域85流出并容许流体从流体管62进入入口区域83。在出口区域85中，流体绕泵体的拐角并沿邻近侧运动，被刷头腔的壁约束到泵和基座元件装配的地方中。流体进入基座元件76中的腔68，绕过靠近腔68端部的小转向元件92进入分配阀70的基部96的内孔94中。

分配阀70如图8所示。内孔94的直径约为0.07英寸。所示实施例中的阀70为0.34英寸高。阀70被称为一种“鸭嘴”阀，这是由于它在其自由端101处具有两个会聚的唇部98和100。这两个唇部98和100通常是关闭的，即被压在一起，以防止流体泄漏和/或倒流。流体由刷头的往复动作及泵的作用移送到阀70中，具有足够的压力强迫所述的两个唇部98和100分开，在喷在活塞的各“排出”冲程中将流体分配到硬毛区域。所示的阀不仅防止作为一种潜在污染源的流体倒流，而且有助于使泵自起动并确定输出流速。

“鸭嘴”阀70是已知的，其由一种弹性塑料材料制成，以便于唇部98和100的打开和关闭。在所示实施例中，阀70的自由端在刷头基座上约伸处0.18英寸。它一般近似设置在硬毛纵向和横向的中部，尽管这对于系统的正确工作是不必要的。

在工作中，当刷头18左右运动以使活塞球在柱状孔75的入口和出口端之间往复运动时，小体积的流体（约0.00013cc）将被依次从储存器通过泵66移送到分配阀70。阀70的唇部98、100与各个小流体浪稍微分开，但在该浪经过之后再靠近。连续量的流体被强迫进入硬毛区域并由移动硬毛的动作输送到牙齿的所需区域。

因此，在硬毛动作的过程中有一个流体到硬毛区域的连续输送（以小的连续喷射）。该系统比已有的系统具有一些优点。第一，流体输送的连续（以连续喷射或脉冲的形式）特性防止流体在刷牙期间变稀，即这种情况，当流体是传统的牙膏或口腔药物时，它在开始刷牙时被引入。第二，应用本发明使所述流体的效果增强，这是由于流体能被直接供给硬毛作用的区域。

在所示牙刷的特殊作用中，以所述方式并具有'590专利所述的包括频率和振幅的工作特性，牙膏或药物流体的作用被反复增强，这是由于硬毛运动所产生的流体的空穴或其它作用。另外，这是流体到嘴部特定区域的精确输送，包括齿龈线中及牙齿之间的牙龈区域。

因此，本发明的流体分配系统不仅实用和可靠，而且当它在选定的工作条件下使用时还具有显著的治疗效果，例如'590专利中牙刷的运动所产生的那些。

如上所述，所示实施例中的储存器50具有一个使用单位的容积，即所述容积足够一次普通长度的刷牙，即两分钟左右。本发明的系统

还包括一个用于一个使用单位储存器的再充系统。该再充系统如图9至14所示。

图9表示整体上的再充装置，其与图1所示的牙刷一起使用。为了再充牙刷头部的一个使用单位的储存器50，再充装置102向下安装在牙刷的顶部，牙刷的头部和刷头部分向上延伸到再充装置102的内部。所述再充装置包括一个壳体104，该壳体具有一个可以示出再充装置一部分内部的可视窗106，因此用户可以确保牙刷的正确插入。

图10表示再充装置的各部分。它包括壳体104和一个内芯元件110。内芯元件110包括一个下部112，该下部的构造可以容纳牙刷头部的上部并与其配合。刷头18向上延伸穿过芯元件110中的一个孔114。芯元件110具有相对的侧轨道116和118，它们在芯元件的外表面119上从芯元件的下边缘117向上延伸。轨道116和118与壳体104内表面120上相应的配合导槽（未示出）配合。

这种设置容许芯元件110以可靠的方式在壳体104中上下运动一定的距离，没有旋转或左右运动。从芯元件110下部112的顶部向上延伸的是一个窄的延伸元件122，在其上端具有一个水平唇部124，在其底部具有两个间隔开的翼形元件126和128，它们邻近所述下部的顶部。芯元件110的下部112在其上端还包括一个邻近翼形元件128的平面部分130。穿过该平面部分130延伸的是一个小孔132。孔132便于容纳再充装置的一个再充针。

位于芯元件110之上的是一个泵装置140，其中包括一个再充储存器111。位于泵装置140之内的是一个填充针142，其与安装到一个密封装置144的针的底部141垂直地取向，密封装置144以一种和泵室内周表面密封的关系装配到泵装置140的泵室部分145。位于泵装置140顶部中的是一个孔150，该孔与泵装置内的再充储存器111连通。一个顶部元件154咬合到泵装置140的上唇部156。在顶部元件154中央处的是一个腔形部分158，在该部分的底部有一个孔160（在图10中局部地示出），该孔可以与泵装置中的孔150对准或不对准，这取决于所述顶部元件的转动位置。当顶部元件处于正确位置以使这两个孔160和150对准时，再充储存器111可以通过这两个孔填充。

图11至13是横剖图，表示用于牙刷中一个使用单位储存器50的再充过程的步骤。在图11中，牙刷刚被设置在再充装置内部，芯元件110

位于其最下位置。在该位置中，延伸元件122的唇部124保留在再充装置内部的一个接收槽170中，和没有牙刷位于再充装置中的情况相同。在该位置中，填充针142的下端164恰好穿过芯元件表面130中的孔132。针底部141及密封装置144位于泵室145内的它们的最下位置。

5 泵室145内的一个弹簧174趋于将针底部141及密封装置144保持在其最下位置。泵室145（原文为144？）下端的室入口阀176位于其关闭位置，如图所示。针142的主体通过密封件178被密封到室145的下边缘元件149。

当牙刷再被用户插入芯元件110中时，牙刷头部的特殊构造的弯曲肩部推靠牙刷元件122底部的翼形元件126、128（图10）。该动作迫使延伸元件122向后足够的距离，以便延伸元件顶部的唇部124移出接收槽170，同时容许芯元件在再充装置中再向上移动。另外由用户作用在牙刷上的向上压力导致芯元件110的进一步向上运动，填充针142进一步穿过芯元件中的孔132并进入牙刷头部14的再充阀56中（图

15 2所示）。

图12表示芯元件运动的一个中间位置，在该处芯元件110的上表面抵靠填充针142的一个肩部180。为了到达该中间位置，芯元件已经在填充装置中向上运动了，但针本身及填充针142的底部141连接的密封装置144都还没有在泵室145中运动。入口阀176保持关闭。

现在参照图13，牙刷的进一步向上运动将导致芯元件110在填充装置中的进一步向上运动，所述芯元件向上推靠在填充针142的肩部180。针142、其底部141及密封装置144在泵室145中向上移动，抵抗弹簧174的作用，在针底部141和泵室145的下端产生一个容积187。来自填充储存器的流体由通过入口阀176的真空作用抽到容积187中，所述阀向上运动。当针底部141及密封装置144抵抗泵室145中弹簧174的作用向上移动时，容积187充满流体。

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填充针142具有沿其周边隔开的孔，在该处它连接底部141。更详细地说，针142通过一种“交叉丝”结构连接底部141，“交叉丝”结构的相邻部分之间的空间通到针的中空内部。当芯元件在填充装置中到达其最高位置时，容积187具有近似4ml的容量而且完全充满来自填充储存器111的流体。

30

在该位置处，刷头稍回退。当牙刷被抽回时，弹簧174的作用迫使密封装置144、针底部141及填充针142本身在泵室145中向下移动，从而减小容积187。容积187中的流体被强迫通过填充针底部的“交叉丝”孔向下进入并通过填充针142的中空内部并从所述针下端的孔188出去。

如上所述，在此位置处的针142中的孔188是位于牙刷的填充阀中。参照图4A和4B，针142向下延伸穿过阀56中的一个顶部密封元件189近似到达其下端190。在区域193中阀56密封地围绕针142，防止流体从针通过排出口194的任何倒流，出口194与流体管路60连接。从所述针中地孔188出来的流体通过口192进入流体管路52（图3），其导向体内储存器50。

从填充装置进入储存器50的流体量在所示实施例中约为4毫升，这是适于一次使用单位的量。但是，如果需要的话，取决于储存器的容量，其它量可以被分配到体内储存器中。当牙刷从填充装置被充分地往回拉时，所述针、其底部及密封装置回到它们在再充装置中的原始位置，如图12所示。然后将牙刷从所述再充装置拆下，准备一次使用流体，即一次刷牙。体内储存器50的再充发生在每次刷牙之前。

当所示实施例包括一个一次使用单位的储存器，而且该储存器正为每次刷牙而被填充时，可以调整储存器以便容纳多于一次使用的流体。但是，外部再充装置对于储存器的方便再充仍是有用的，容许使用一个位于牙刷体内的较小流体储存器，因此节约了空间和泵送需求。

至此，已经描述了用于电动牙刷的一种新的流体分配和再充系统。该系统包括位于牙刷可动头部中的一个小的使用单位储存器。一个流体管路从所述储存器延伸到位于刷头中的一个小泵元件中。该泵元件随着刷头的往复运动提供了流体从储存器到牙刷头中一个分配阀的运动，从而使流体移动到硬毛。

体内储存器在每次使用之前由一个单独的再充装置再充，其中牙刷被插入该装置。以特定顺序的刷头相对于再充装置的运动导致再充储存器中的流体通过牙刷头部中的一个填充针和再充阀进入刷体内的一个使用单位的储存器。

图5至8所述的泵结构的一个替换实施例见图15至20。被标以202的整个泵局部地形成在刷头204的底部203中。一个盖元件214也是整体泵202的一部分。流体，例如一种牙膏通过一个进入管206被提供给泵202。形成在底部203中的泵202的一部分是一个圆柱状的泵孔208。
5 可在泵孔208中移动的是一个球形活塞元件210，该元件也被称为活塞球。

位于泵孔208排出端226处的是一个塞子元件212，其在泵孔208中具有一个特定的构造。一个浅的流体出口部分215从泵孔208的排出端226延伸到一个单向阀216，该阀将流体引导至硬毛。图15至20所示
10 的替换实施例通常更有效而且会比前述实施例产生更高的压力。

现在参照图15至18，入口管206通过刷头204的邻近端219提供流体，例如牙膏到一个短通道220，该通道被模制到刷头（图16）中并将管206的远端222连接到一个入口连接件224，该连接件在入口端227
15 连接到泵孔208。在所示实施例中，入口连接件224被设置在泵孔208的顶部。

所示实施例中的泵孔208直径比球形活塞元件210的直径稍大，在所示实施例中为2mm。仅需要使泵孔的直径稍大于活塞元件210的直径，以便当活塞元件在所述孔中移动以从泵孔中排出流体时维持泵孔中的压力并使通过活塞元件的流体泄漏非常小。在所示实施例中，泵
20 孔的直径约为2.1mm、长度约为4.5mm。

设置在泵孔208排出端226的是一个塞子212。塞子212具有一个内表面232（朝向泵孔），该表面是中凹的，即盘形的，这在图20中看得最清楚。在所示实施例中，所述弯曲稍大于球形活塞元件的弯曲度。在所示实施例中，塞子由弹性材料制成。

在球形活塞元件在泵孔208中运动期间，它在流体排出冲程的末尾接触表面232，期间流体被活塞元件的运动从泵孔的入口端推到出口端。在所示实施例中，所述活塞元件由碳化钨材料制成。例如，碳化钨的增加密度可认为是相对于钢材增加了泵的压力和性能。

影响性能的泵变量包括球形活塞元件的直径、球形活塞的横截面积及球形活塞的质量。活塞元件和孔之间的间隙也影响性能。泵孔
30 中的流体通过出口234排出，该出口位于泵孔排出端的顶部（图20）。

从泵孔208排出的流体直接进入流体出口部分215, 该部分在所示实施例中约为0.25英寸长、0.10英寸宽、0.02英寸高, 并被限定在刷头和盖元件214的下表面214a中。盖元件214从管206的通道220的起始处延伸到经过流体排出通道215的端部229的一位置。

5 现在参照图19, 由盖214下表面214a限定的流体排出部分包括两个翼形延伸件236和238, 它们从流体排出通道的纵向侧边朝向彼此延伸, 使那个位置处的排出通道变窄。这两个延伸件236和238通常是三角状, 而且经过约0.08英寸的距离后将所述通道的宽度从0.10英寸左右缩小到0.04英寸左右。在将从泵孔208排出的流体向刷头中的排出
10 孔引导中, 两个延伸件236和238用作一个流体二极管, 沿排出方向提供的流体流动阻力比沿相反方向提供的小。在延伸件236和238之外的是一个元件240, 该元件居中地设置在排出通道的两侧之间并从盖的下表面214a向下垂入排出通道。所述排出通道终止在其侧面向内倾斜处并汇合在一个弯曲部分中。

15 直接位于元件240下方模制到刷头底部203中的的是一个圆形孔244, 所示实施例中该孔的直径约为0.10英寸。孔244通过刷头延伸, 通到硬毛延伸的表面。设置在孔244中以便它离开刷头表面延伸并位于硬毛之间的是排出单向阀216。在所示实施例中, 单向阀216是楔形的, 与图5至8中的鸭嘴形结构相对。总得来说, 如上所示, 图15至20
20 的泵实施例通常更有效而且比前面披露的泵产生更好的压力。

尽管本发明的优选实施例在此处是被作为示例的, 但应该理解在不偏离本发明精神的情况下该实施例可以包含各种改变、变型及替换, 这由所附的权利要求限定。

说明书附图

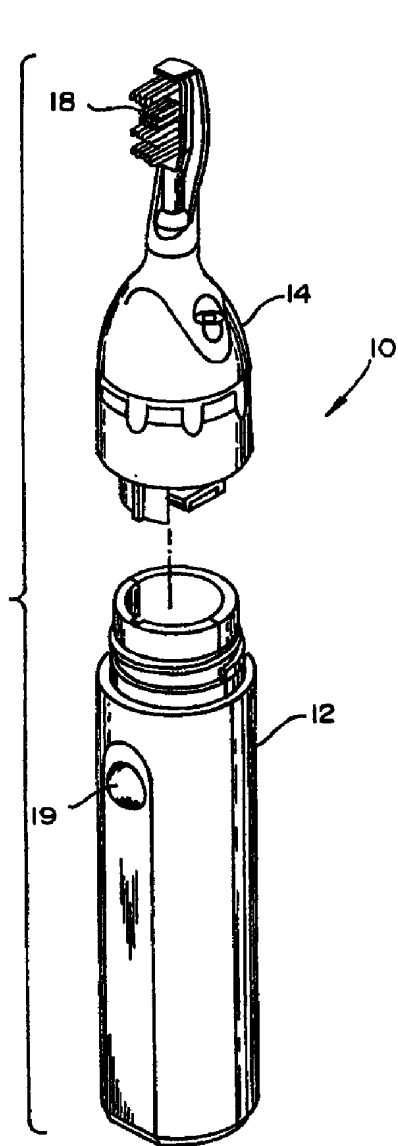


图 1

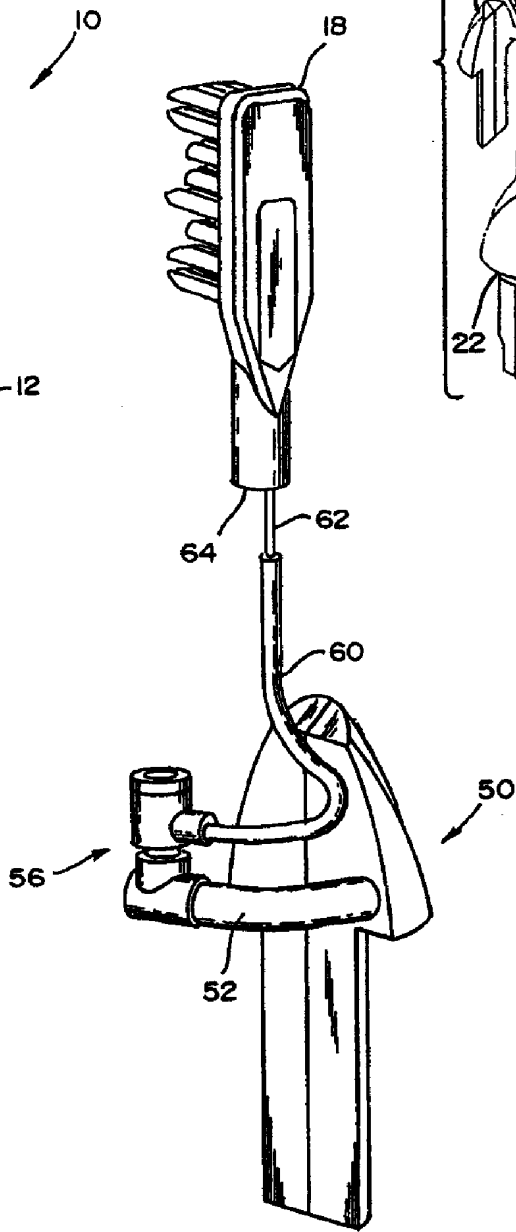


图 3

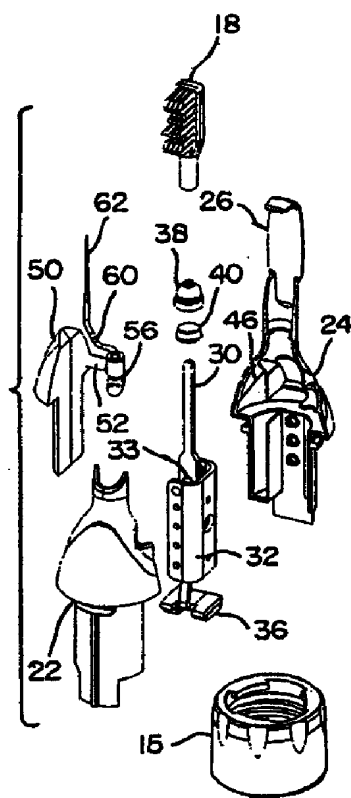


图 2

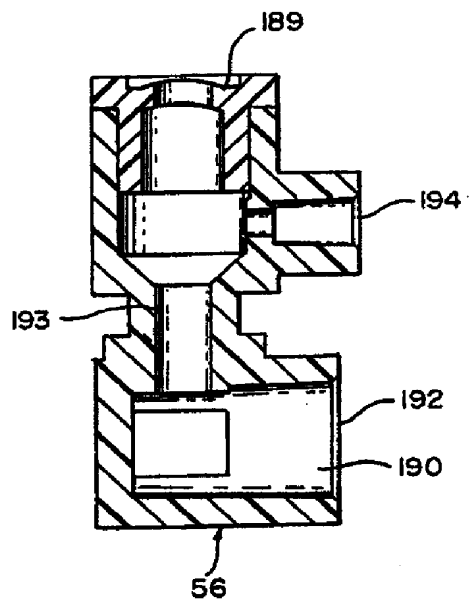


图 4A

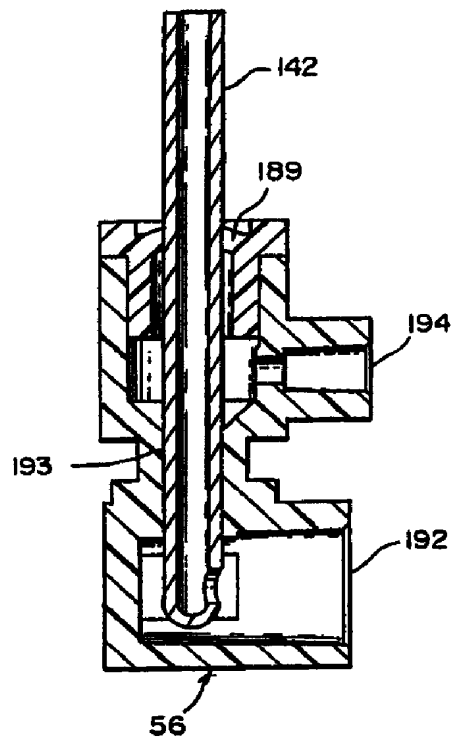


图 4B

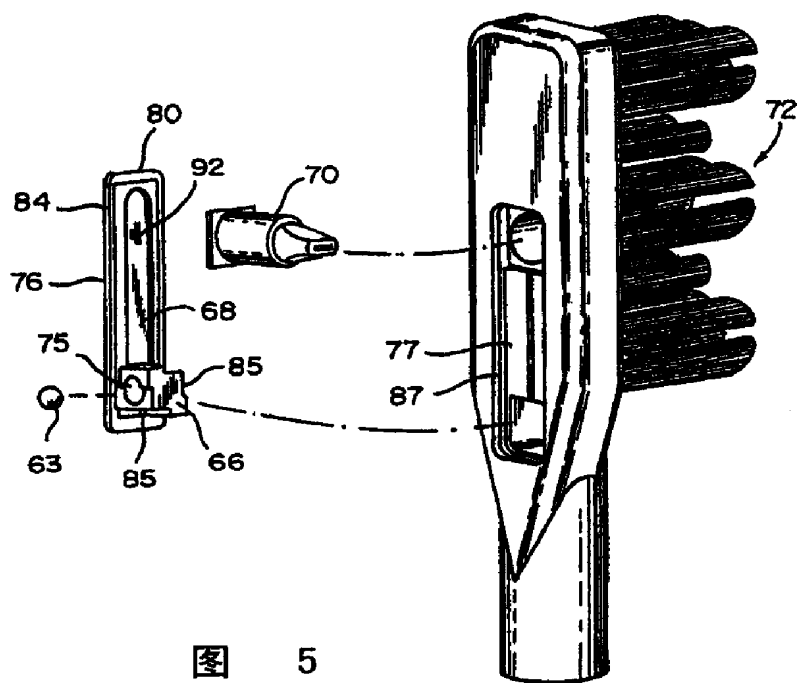


图 5

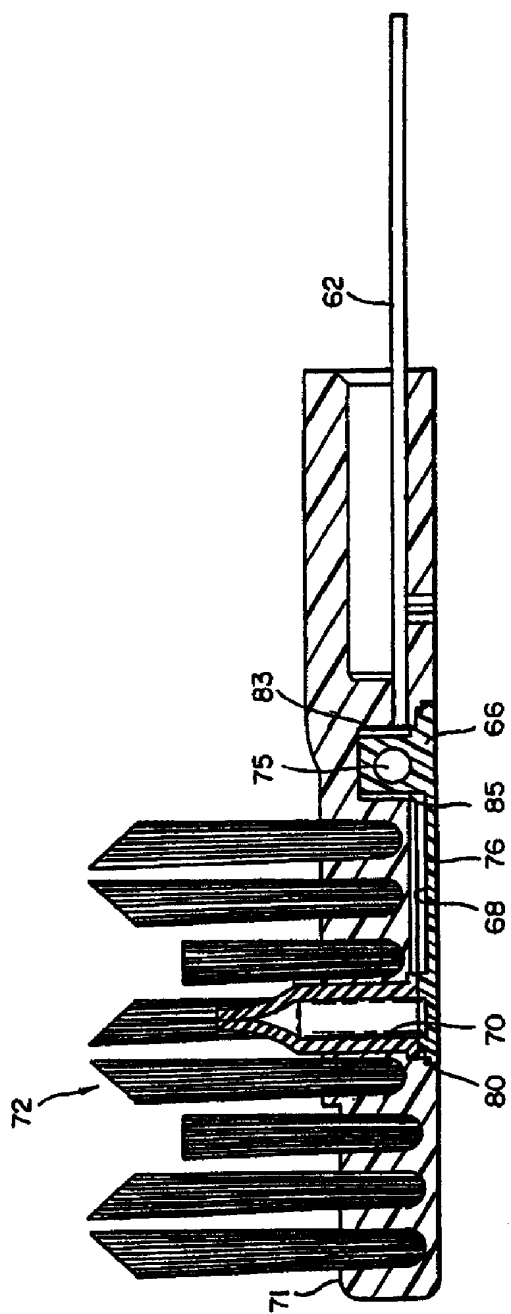


图 6

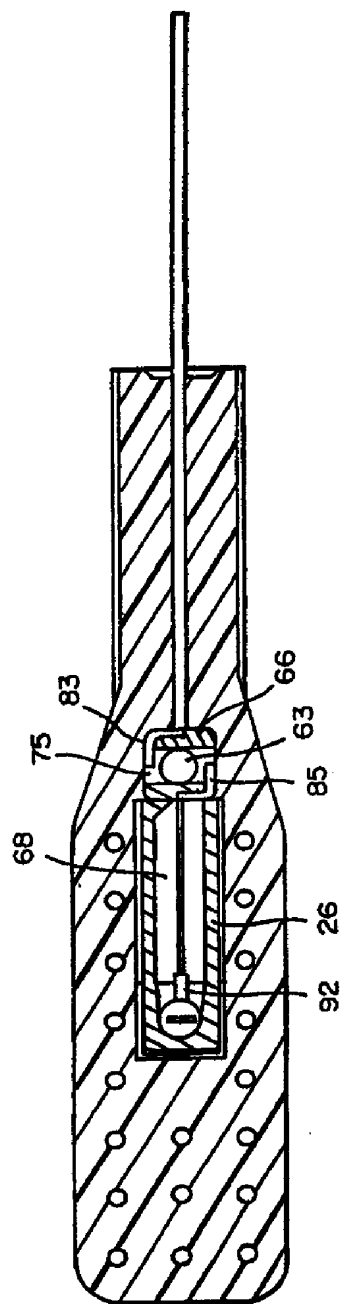


图 7

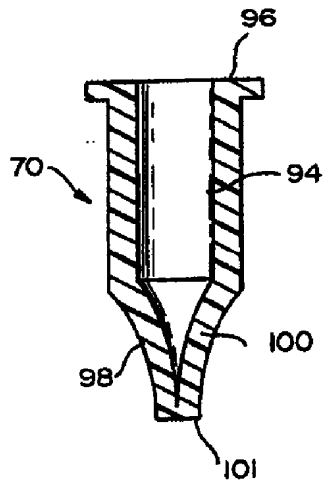


图 8

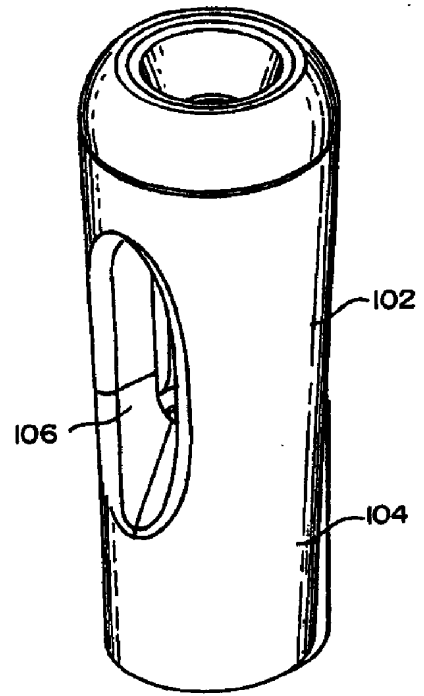


图 9

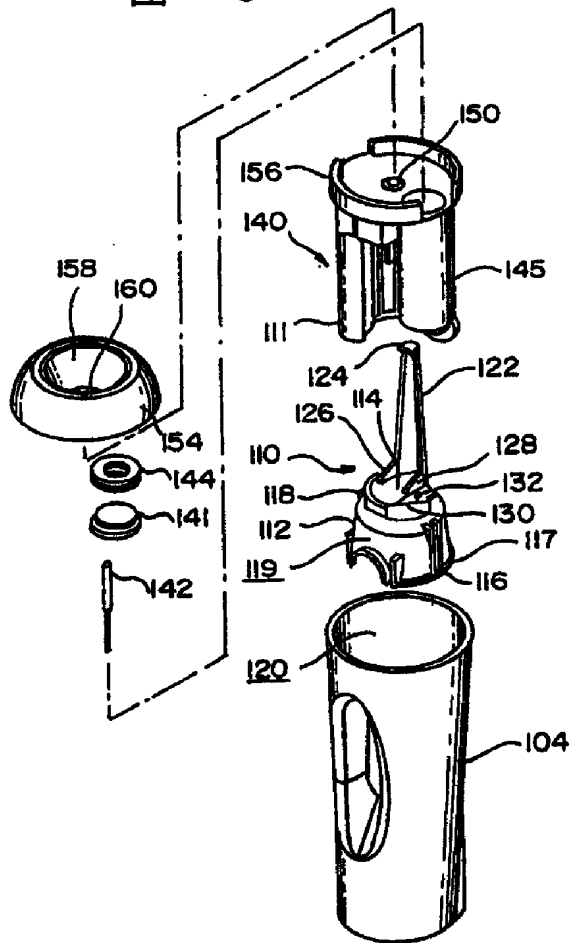


图 10

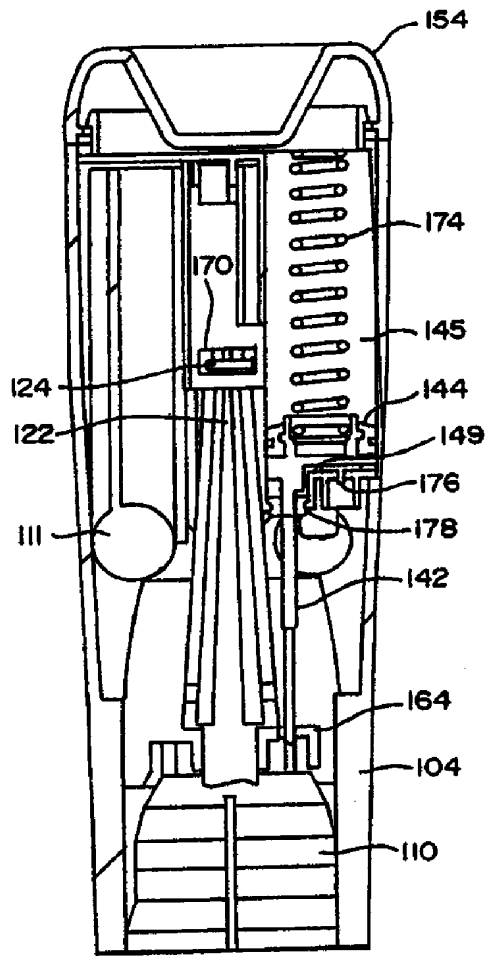


图 11

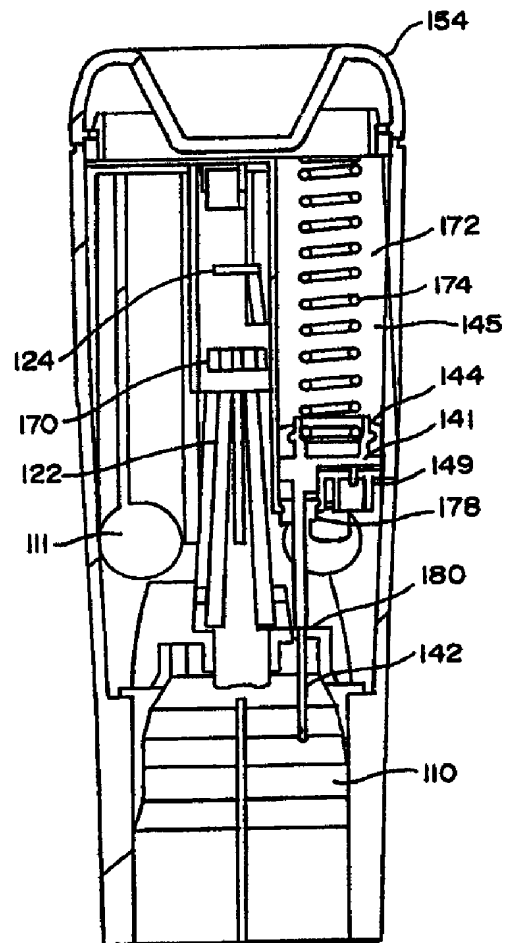


图 12

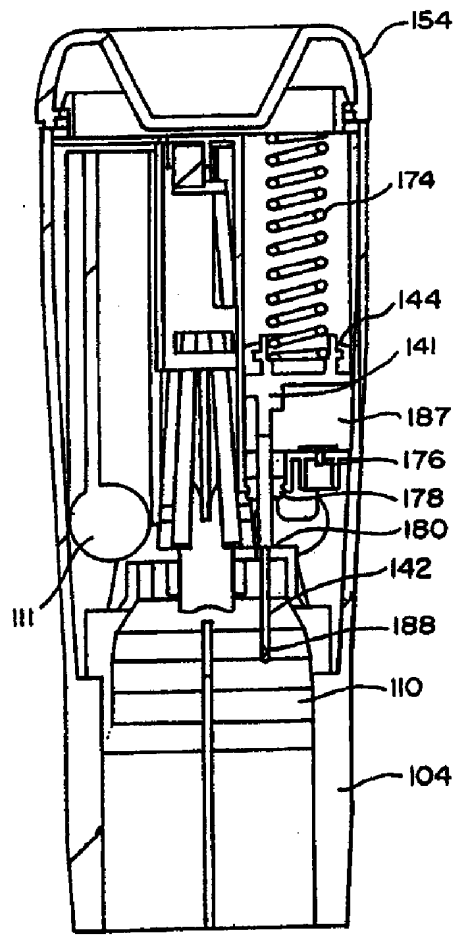


图 13

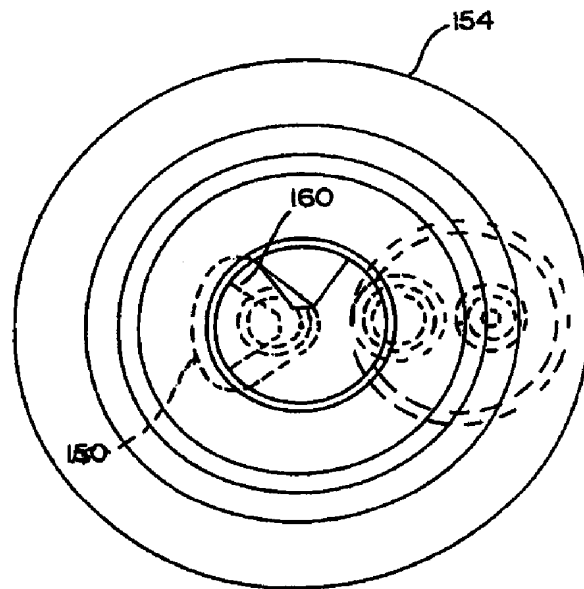


图 14

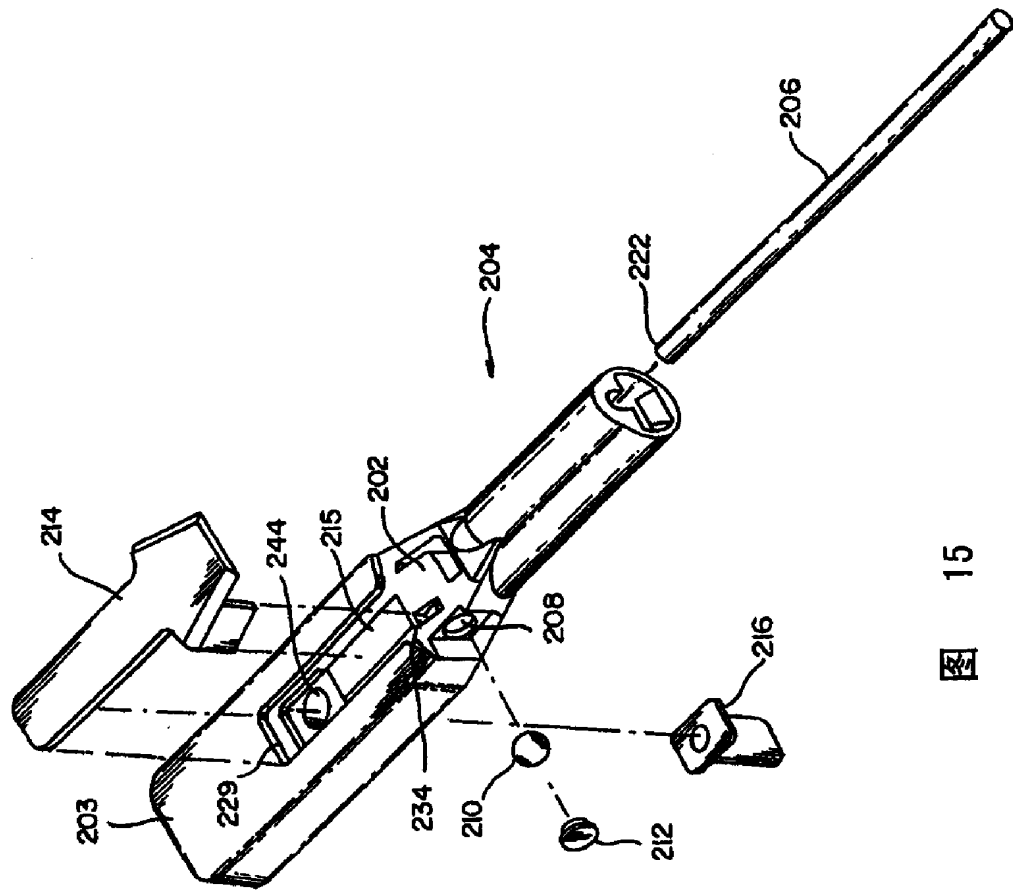


图 15

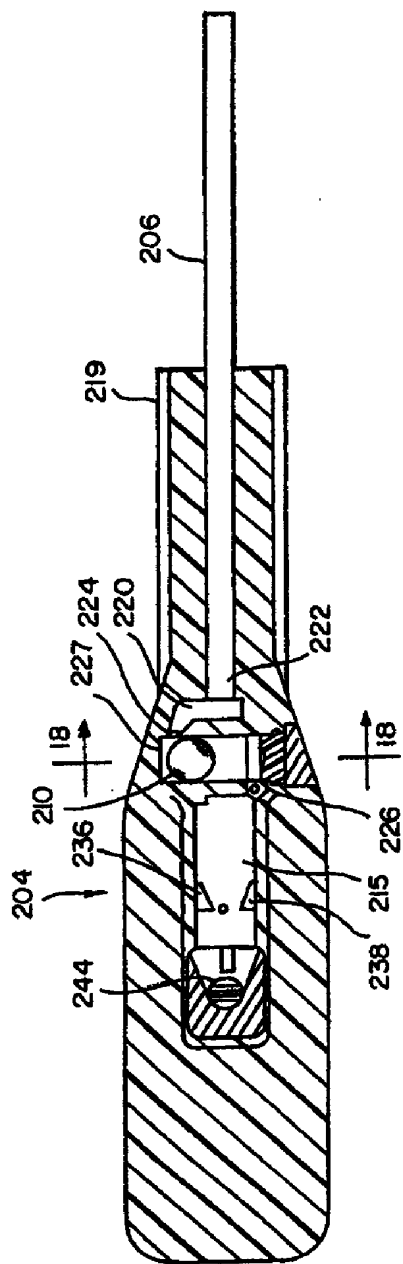


图 16

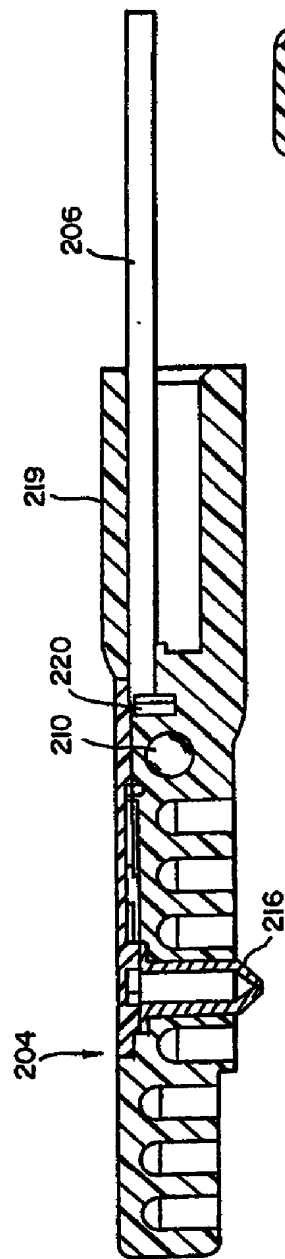


图 17

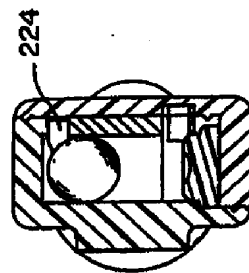


图 18

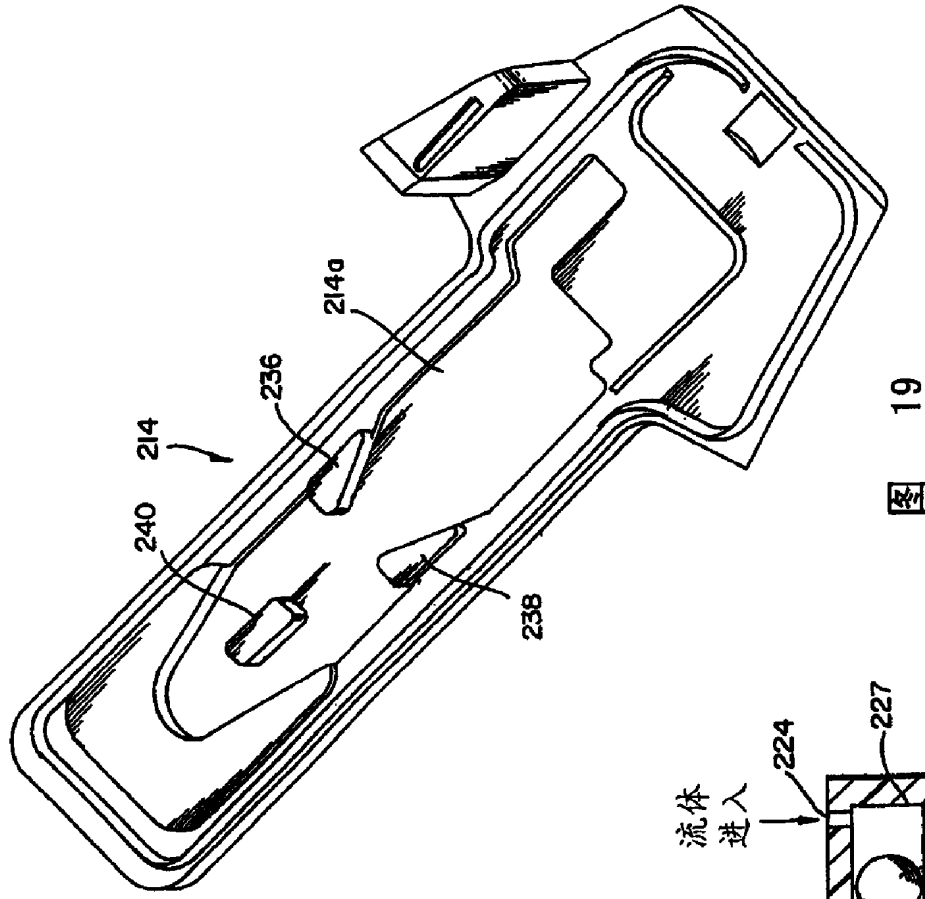


图 19

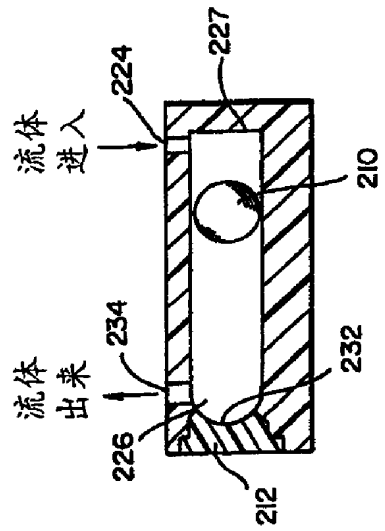


图 20

Cooling system for photocosmetic device**Publication number:** CN1535126 (A)**Publication date:** 2004-10-06**Inventor(s):** ALTSHULER G B [US]; CARUSO J D [US]; ZENZIE H M [US]**Applicant(s):** PALOMAR MEDICAL TECH INC [US]**Classification:**






- **international:** **A61B17/00; A61B18/20; A61F7/00; A61N5/06; A61B18/00; A61N5/00; A61B17/00; A61B18/20; A61F7/00; A61N5/06; A61B18/00; A61N5/00; (IPC1-7): A61B18/18**

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Application number: CN20028014860 20020523

Priority number(s): US20010292827P 20010523; US20020052474 20020118;
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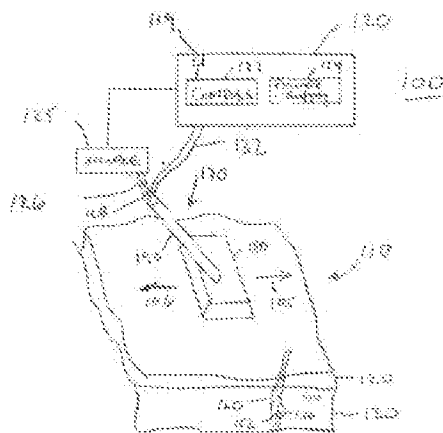
 CN1262249 (C)
 WO02094116 (A1)
 WO02094116 (A9)
 JP2009101183 (A)
 JP2004527330 (T)

more >>

Abstract not available for CN 1535126 (A)

Abstract of corresponding document: **WO 02094116 (A1)**

Photocosmetic device (100) for use in medical or non-medical environments (e.g., a home, barbershop, or spa), which can be used for a variety of tissue treatments. Radiation is delivered to the tissues via optical systems (520 for example) designed to pattern the radiation and project the radiation to a particular depth. The device has a variety of cooling systems including phase change cooling solids and liquids to cool treated skin and the radiation sources (510 for example). Contact sensors (1712) and motion sensor (1820) may be used to enhance treatment. The device may be modular to facilitate manufacture and replacement of parts.



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[11] 公开号 CN 1535126A

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[32] 2002. 3. 12 [33] US [31] 60/363,798

[32] 2002. 3. 12 [33] US [31] 60/363,871

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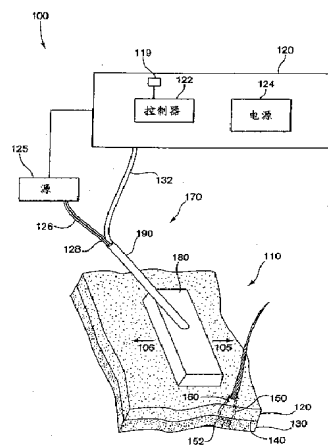
代理人 崔幼平

权利要求书 6 页 说明书 42 页 附图 33 页

[54] 发明名称 用于光美容装置的冷却系统

[57] 摘要

光美容装置(100)用于医学或非医学环境(例如家庭、理发店或者矿泉疗养地)中,可用于进行多种组织处理。射线通过设计用于对射线定型并将射线投射至特定深度的光学系统(例如 520)输送至组织。这种装置具有多种包括相变冷却固体和液体的冷却系统以便冷却受到处理的皮肤和射线源(例如 510)。接触传感器(1712)和运动传感器(1820)可以用来加强处理效果。这种装置可为模块式以便于制造和更换零件。



1. 一种用于病人皮肤的一个区域上的光美容装置，包括：
一紧靠病人皮肤使用的处理头部；
5 至少一个置于处理头部内并且适于将射线投射于皮肤区域上的电磁射线源；
一与该至少一个源保持热联接的冷却表面；以及
一用于将相变物质引导至冷却表面上的机构。
2. 根据权利要求 1 所述的装置，其特征在于，相变物质包括一
10 液体。
3. 根据权利要求 1 所述的装置，其特征在于，相变物质包括一固体。
4. 根据权利要求 1 所述的装置，其特征在于，冷却表面具有纹理。
- 15 5. 根据权利要求 4 所述的装置，其特征在于，纹理为线性凹槽型式。
6. 根据权利要求 4 所述的装置，其特征在于，纹理为同心凹槽型式。
7. 根据权利要求 4 所述的装置，其特征在于，纹理为多个突起。
- 20 8. 根据权利要求 2 所述的装置，其特征在于，机构包括一喷口。
9. 根据权利要求 8 所述的装置，其特征在于，机构还包括一联接至喷口上的阀，其中阀控制着投射于冷却表面上的液体量。
10. 根据权利要求 9 所述的装置，还包括一用于产生显示至少一部分皮肤区域的温度的信号的热传感器和一用于接收来自热传感器的
25 信号并响应于温度而控制着阀的控制器。
11. 根据权利要求 1 所述的装置，还包括一用于容放物质的容器，容器联接至机构上。
12. 根据权利要求 2 所述的装置，其特征在于，液体包括一致冷剂。
- 30 13. 根据权利要求 12 所述的装置，其特征在于，致冷剂包括四氟代乙烷。
14. 根据权利要求 3 所述的装置，其特征在于，固体包括冰。

15. 根据权利要求 3 所述的装置, 其特征在于, 固体包括一有机化合物。

16. 根据权利要求 3 所述的装置, 其特征在于, 固体为镓/铟合金。

5 17. 根据权利要求 1 所述的装置, 其特征在于, 冷却表面为一向源提供电力的导热电极的一个表面。

18. 根据权利要求 1 所述的装置, 其特征在于, 冷却表面为一与源保持热联接的导热散热片的一个表面。

10 19. 根据权利要求 2 所述的装置, 其特征在于, 该至少一个源具有一定长度, 并且冷却表面具有至少一用于接收相变物质的通过其中的通道。

20. 根据权利要求 2 所述的装置, 其特征在于, 该至少一个源具有一定长度, 冷却表面具有多个用于接收相变物质的通过其中的通道, 这些通道沿着长度方向对齐。

15 21. 一种用于病人皮肤的一个区域上的光美容装置, 包括:

一紧靠病人皮肤使用的处理头部;

至少一位于处理头部内并且适于将射线通过处理头部投射于皮肤区域上的电磁射线源; 以及

20 一联接于处理头部上并适于将第一物质投射至病人皮肤上的第一机构。

22. 根据权利要求 21 所述的装置, 还包括一用于将射线传送至皮肤区域的光学系统, 这种光学系统具有一适于与病人皮肤相接触的表面。

25 23. 根据权利要求 21 所述的装置, 还包括一与该至少一个源和所述表面保持热联接的冷却表面, 以及一用于将相变物质投射于冷却表面上的第二机构, 其中第一机构适于使用通过相变物质的相变而形成的气体来将第一物质驱至病人皮肤上。

30 24. 根据权利要求 21 所述的装置, 还包括一与源和所述表面保持热联接的冷却表面, 以及一适于将第一物质的第一部分投射于冷却表面上的第二机构。

25. 根据权利要求 24 所述的装置, 其特征在于, 第一物质为一液体而投射于皮肤上的第一物质的第二部分为通过第一物质的相变而

产生的气体。

26. 根据权利要求 24 所述的装置, 其特征在于, 第一物质为一固体而投射于皮肤上的第一物质的第二部分为通过第一物质的相变而产生的液体。

5 27. 根据权利要求 24 所述的装置, 其特征在于, 第一物质为一固体而投射于皮肤上的第一物质的第二部分为通过第一物质的相变而产生的气体。

28. 根据权利要求 21 所述的装置, 其特征在于, 第一物质包括一液体。

10 29. 根据权利要求 28 所述的装置, 其特征在于, 液体包括一洗液。

30. 根据权利要求 21 所述的装置, 其特征在于, 第一物质包括一气体。

15 31. 根据权利要求 30 所述的装置, 其特征在于, 气体包括已冷却过的空气。

32. 根据权利要求 21 所述的装置, 其特征在于, 第一物质包括多种组分。

33. 根据权利要求 23 所述的装置, 其特征在于, 冷却表面为一向源提供电力的导热电极的一个表面。

20 34. 根据权利要求 23 所述的装置, 其特征在于, 冷却表面为一与源保持热联接的导热散热片的一个表面。

35. 根据权利要求 21 所述的装置, 其特征在于, 源为二极管激光棒、发光二极管和灯之一。

25 36. 一种用于病人皮肤的一个区域上的装置, 包括:
一紧靠病人皮肤使用的处理头部;

至少一置于处理头部内并且适于将电磁射线投射于皮肤区域上的电磁射线源;

一与该至少一个电磁射线源保持热联接并且包括至少一通过其中的通道的冷却表面; 以及

30 一用于将物质投射至冷却表面上以及该至少一通道中的机构。

37. 根据权利要求 36 所述的装置, 其特征在于, 物质为一液体。

38. 根据权利要求 36 所述的装置, 其特征在于, 物质为一气体。

39. 一种用于病人皮肤的一个区域上的装置, 包括:
至少一适于将射线投射于皮肤区域上的电磁射线源;
一与该至少一个源保持热联接的冷却表面; 以及
一与冷却表面保持热联接的固体块, 固体块随着从冷却表面上吸
5 热而发生相变。
40. 根据权利要求 39 所述的装置, 其特征不在于, 固体块为冰。
41. 根据权利要求 39 所述的装置, 其特征不在于, 固体块为干冰。
42. 根据权利要求 39 所述的装置, 还包括一带动固体块与冷却
表面形成接触的机构。
- 10 43. 根据权利要求 39 所述的装置, 还包括一处理头部, 其中源
置于该处理头部中。
44. 根据权利要求 39 所述的装置, 其特征不在于, 源为二极管激
光棒、发光二极管和灯之一。
45. 根据权利要求 39 所述的装置, 其特征不在于, 冷却表面为一
15 向源提供电力的导热电极的一个表面。
46. 根据权利要求 39 所述的装置, 其特征不在于, 冷却表面为一
与源保持热联接的导热散热片的一个表面。
47. 一种用于病人皮肤的一个区域上的装置, 包括:
至少一适于将电磁射线投射于皮肤区域上的电磁射线源;
20 一与该至少一个源保持热联接的冷却表面;
一与冷却表面保持热联接的固体块, 固体块的至少一部分随着从
冷却表面上吸热而变为液体; 以及
一适于接收液体的一部分并且将这部分液体投射于病人皮肤上的
排放孔。
- 25 48. 根据权利要求 48 所述的装置, 还包括一用于将液体与一化
学物质组合在一起并将液体和化学品的组合引导至病人皮肤上的机
构。
49. 一种用于病人皮肤的一个区域上的装置, 包括:
至少一适于将电磁射线投射于皮肤区域上的电磁射线源;
30 一与该至少一个源保持热联接的冷却表面;
一与冷却表面保持热联接并且包含至少一第一化学化合物和一第
二化学化合物的反应室, 第一和第二化学化合物经过选择以便在反应

室内提供吸热反应。

50. 根据权利要求 49 所述的装置, 其特征在于, 冷却表面为一向源提供电力的导热电极的一个表面。

51. 根据权利要求 49 所述的装置, 其特征在于, 冷却表面为一
5 与源保持热联接的导热散热片的一个表面。

52. 一种用于病人皮肤的一个区域上的装置, 包括:

一紧靠病人皮肤使用的处理头部;

至少一置于处理头部内并且适于将电磁射线投射于皮肤区域上的
电磁射线源; 以及

10 一与该至少一个电磁射线源保持热联接的冷却表面, 冷却表面具有一通过其中的通道, 用于容许一低沸点液体流动至冷却表面的一个表面上。

53. 根据权利要求 52 所述的装置, 还包括一联接至该通道的阀, 用于控制低沸点液体的蒸发。

15 54. 根据权利要求 53 所述的装置, 还包括一用于产生显示皮肤区域的温度的信号的热传感器和一用于接收来自热传感器的信号并响应于信号控制着阀的控制器。

55. 根据权利要求 52 所述的装置, 其特征在于, 一压力源联接至通道上以便控制低沸点液体的沸腾情况。

20 56. 根据权利要求 52 所述的装置, 其特征在于, 源为激光二极管棒、发光二极管和灯之一。

57. 一种用于病人皮肤的一个区域上的装置, 包括:

一紧靠病人皮肤使用的处理头部;

25 至少一置于处理头部内并且适于将射线投射于皮肤区域上的电磁射线源;

一与该至少一个源保持热联接的热传播器; 以及

一与热传播器保持热联接的冷却表面。

58. 根据权利要求 57 所述的装置, 其特征在于, 源为二极管激光棒、发光二极管和灯之一。

30 59. 根据权利要求 57 所述的装置, 其特征在于, 冷却表面为一向源提供电力的导热电极的一个表面。

60. 根据权利要求 57 所述的装置, 其特征在于, 冷却表面为一

与源保持热联接的导热散热片的一个表面。

61. 一种用于冷却热产生装置的冷却系统，包括：

一与热产生装置保持热联接的冷却表面；

5 一适于投射高压液体的喷嘴，液体形成了在冷却表面上流动的液体。

62. 根据权利要求 61 所述的冷却系统，其特征在于，这种高压液体在位于喷嘴与冷却表面之间的整个距离上形成一流。

63. 根据权利要求 61 所述的冷却系统，其特征在于，冷却表面带有纹理。

10 64. 根据权利要求 61 所述的冷却系统，还包括一用于将液体重新引回至冷却表面上的冷却室。

65. 根据权利要求 64 所述的冷却系统，其特征在于，冷却室包括侧壁和一盖。

用于光美容装置的冷却系统

发明背景

5 相关申请

本申请对 2002 年 3 月 12 日提交的临时申请序号 60/363798 的优先权提出要求。本申请还是 2002 年 1 月 18 日提交的申请序号 10/052,474 的部分继续, 而该申请为 1999 年 12 月 28 日提交的申请序号 09/473,910 的继续, 而该申请对 1999 年 1 月 8 日提交的临时申请序号 60/115,447 的优先权提出了要求, 对 1999 年 11 月 9 日提交的临时申请序号 60/164,492 的优先权提出了要求, 并且是 1998 年 5 月 13 日提交的申请序号 09/078,055 即现在的美国专利 U. S. Patent No. 6,273,884 的部分继续, 而该申请对 1997 年 5 月 15 日提交的临时申请序号 60/046,542 和 1998 年 3 月 12 日提交的临时申请序号 60/077,726 的优先权提出了要求。本申请还是 1999 年 3 月 12 日提交的申请序号 09/268,433 的部分继续, 而该申请对 1999 年 1 月 8 日提交的临时申请序号 60/115,447 和 1999 年 1 月 8 日提交的临时申请序号 60/077,794 的优先权提出了要求, 并且是 1996 年 12 月 2 日提交的申请序号 08/759,036 即现在的美国专利 U. S. Patent No. 6,015,404 的部分继续, 并且是现在已放弃的 1996 年 12 月 2 日提交的申请序号 08/759,136 的部分继续, 并且是 1998 年 5 月 13 日提交的申请序号 09/078,055 即现在的美国专利 U. S. Patent No. 6,273,884 的部分继续, 而该申请对 1997 年 5 月 15 日提交的临时申请序号 60/046,542 和 1998 年 3 月 12 日提交的临时申请序号 60/077,726 的优先权提出了要求。本申请还是 2000 年 8 月 9 日提交的申请序号 09/634,981 的部分继续, 而该申请是 1998 年 5 月 13 日提交的申请序号 09/078,055 即现在的美国专利 U. S. Patent No. 6,273,884 的继续, 而该申请对 1997 年 5 月 15 日提交的临时申请序号 60/046,542 和 1998 年 3 月 12 日提交的临时申请序号 60/077,726 的优先权提出了要求。本申请还是 2001 年 4 月 30 日提交的申请序号 09/847,043 的部分继续, 而该申请对 2000 年 4 月 28 日提交的临时申请序号 60/200,431 的优先权提出了要求。本申请还对 2001 年 5 月 23 日提交的临时申请序号 60/292,827 的优先权提出了要求。本申请还对

2002年3月12日提交的临时申请序号60/363,871的优先权提出了要求。所有这些前面的申请说明书的内容在此引入作为参考。

背景技术

存在多种可使用光美容程序（本文中也可称作光美容处理）来处理的情况，包括光基（例如使用激光器或灯）除毛、各种皮肤病变的处理，纹身去除、面部重修面以及皮肤恢复活力处理。目前，光美容程序利用对位于病人皮肤的表皮/真皮中的目标结构进行破坏性加热的专业级装置来执行。

到目前为止，光美容程序都是在皮肤科医生的办公室里进行，这部分是由于用于执行这些程序的装置价格昂贵，部分是由于与装置相关的安全考虑，以及部分是由于需要对光在病人皮肤上产生的伤口进行护理。这种伤口可能由于大功率射线对病人表皮造成的损坏而产生并且可能会造成显著的疼痛和/或造成感染的危险。尽管某些光美容程序，例如CO₂激光器面部重修面，出于医学原因（例如需要进行操作后的伤口护理）将可以继续皮肤科医生的办公室里执行，但是如果消费者能够按照安全有效的方式执行程序，则大量的光美容程序可以在非医学环境（例如家庭、理发店或者矿泉疗养地）中执行。即使对于在医学环境中执行的程序，减少皮肤损坏也将会缩短恢复时间。

用于医学或非医学环境中的光美容装置可以受益于以下特征。

（1）这种装置必须安全。例如，必须避免造成眼睛和皮肤损伤。（2）优选地，这种装置易于使用，从而使得操作者在只需阅读了简短的训练周期之后就能够实现可接受的美容结果。（3）优选地，这种装置足够结实耐用以便能承受随便滥用。（5）优选地，这种装置易于维护。（6）优选地，这种装置可大量制造。（7）优选地，这种装置可以合理的价格买到。（8）优选地，这种装置很小并且易于存储，例如存储于浴室中。目前可买到的光美容装置在关于一项或更多项以上提出的问题方面具有局限性。

发明内容

本发明的第一方面为一种用于病人皮肤的一个区域上的光美容装置，包括一紧靠病人皮肤使用的处理头部、至少一置于处理头部内并且适于将射线投射于皮肤区域上的电磁射线源、一与该至少一个源保持热联接的冷却表面以及一用于将相变物质引导至冷却表面上的机

构。任选地，相变物质包括一液体。另外，相变物质也可包括一固体。

在第一方面的一些实施例中，表面具有纹理。纹理可为线性凹槽型式或者同心凹槽型式。另外，纹理也可为多个突起。该机构可为一喷口。机构可还包括一联接至喷口上的阀，其中阀控制着投射于冷却表面上的液体量。可使用一热传感器来产生显示至少一部分皮肤区域的温度的信号，并且可使用一控制器来接收来自热传感器的信号并响应于温度而控制着阀。

一容器可包括于内以用于容放物质。在一些实施例中，这种物质为致冷剂。例如，致冷剂包括四氟代乙烷。固体可为冰或者一有机化合物、或者为一镓/铟合金。

冷却表面可为一向源提供电力的导热电极。另外，冷却表面也可为一与源保持热联接的导热散热片的一个表面。冷却表面可具有至少一用于接收相变物质的通过其中的通道。另外，冷却表面也可具有多个用于接收相变物质的通过其中的通道，这些通道沿着长度方向对齐。

本发明的第二方面为一种用于病人皮肤的一个区域上的光美容装置，包括一紧靠病人皮肤使用的处理头部、至少一适于将射线通过处理头部投射于皮肤区域上的电磁射线源、以及一联接于处理头部上并适于将第一物质投射至病人皮肤上的第一机构。电磁射线源可置于处理头部内。这种装置可包括一用于将射线传送至皮肤区域的光学系统，这种光学系统具有一适于与病人皮肤相接触的表面。这种装置可还包括一与该至少一个源和所述表面保持热联接的冷却表面；以及一用于将相变物质投射于冷却表面上的第二机构，其中第一机构适于使用通过第二物质的相变而形成的气体来将第一物质驱至病人皮肤上。这种装置可还包括一与源和所述表面保持热联接的冷却表面，以及一适于将第一物质的一部分投射于冷却表面上的第二机构。

第一物质可为一液体而投射于皮肤上的第一物质的部分为通过第一物质的相变而产生的气体。另外，第一物质也可为一固体而投射于皮肤上的第一物质的部分为通过第一物质的相变而产生的液体。在另外一个替代方案中，第一物质为一固体而投射于皮肤上的第一物质的部分为通过第一物质的相变而产生的气体。

第一物质可为一液体，而这种液体可为洗液。另外，第一物质可

为一气体，而这种气体可为已冷却过的空气。第二物质可包括多种组分。冷却表面可为一向源提供电力的导热电极的一个表面。冷却表面可为一与源保持热联接的导热散热片的一个表面。任选地，源可为二极管激光棒、发光二极管和灯之一。

5 本发明的第三方面为一种用于病人皮肤的一个区域上的装置，包括一紧靠病人皮肤使用的处理头部、至少一置于处理头部内并且适于将电磁射线投射于皮肤区域上的电磁射线源、一与该至少一个电磁射线源保持热联接并且包括至少一通过其中的通道的冷却表面、以及一用于将物质投射至冷却表面上以及该至少一通道中的机构。

10 这种物质可为一液体或一气体。

本发明的第四方面为一种用于病人皮肤的一个区域上的装置，包括至少一适于将射线投射于皮肤区域上的电磁射线源、一与该至少一个源保持热联接的冷却表面以及一与冷却表面保持热联接的固体块，固体块随着从冷却表面上吸热而发生相变。

15 在一些实施例中，固体块为冰或者可为干冰。这种装置可还包括一带动固体块与冷却表面形成接触的机构。这种装置可还包括一处理头部，其中源置于该处理头部中。源可为二极管激光棒、发光二极管和灯之一。

20 冷却表面可为一向源提供电力的导热电极的一个表面或者一与源保持热联接的导热散热片。

本发明的第五方面为一种用于病人皮肤的一个区域上的装置，包括至少一适于将电磁射线投射于皮肤区域上的电磁射线源、一与该至少一个源保持热联接的冷却表面、一与冷却表面保持热联接的固体块、以及一适于接收液体的一部分并且将这部分液体投射于病人皮肤
25 上的排放孔，其中固体块的至少一部分随着从冷却表面上吸热而变为液体。

这种装置可还包括一用于将液体与一化学物质组合在一起并将液体和化学品的组合引导至病人皮肤上的机构。

30 本发明的第六方面为一种用于病人皮肤的一个区域上的装置，包括至少一适于将电磁射线投射于皮肤区域上的电磁射线源、一与该至少一个源保持热联接的冷却表面以及一与冷却表面保持热联接并且包含至少一第一化学化合物和一第二化学化合物的反应室，第一和第二

化学化合物经过选择以便在反应室内提供吸热反应。

冷却表面可为一向源提供电力的导热电极的一个表面，或者冷却表面可为一与源保持热联接的导热散热片的一个表面。

5 本发明的第七方面为一种用于病人皮肤的一个区域上的装置，包括一紧靠病人皮肤使用的处理头部、至少一置于处理头部内并且适于将电磁射线投射于皮肤区域上的电磁射线源、以及一与该至少一个电磁射线源保持热联接的冷却表面，冷却表面具有一通过其中的通道，用于容许一低沸点液体流动至冷却表面的一个表面上。

10 这种装置可还包括一联接至该通道的阀，用于控制低沸点液体的蒸发。这种装置可另外还包括一用于产生显示皮肤区域的温度的信号的热传感器和一用于接收来自热传感器的信号并响应于信号控制着阀的控制器。这种装置可具有一联接至通道以便控制低沸点液体的沸腾情况的压力源。这种源为激光二极管棒、发光二极管和灯之一。

15 本发明的第八方面为一种用于病人皮肤的一个区域上的装置，包括一紧靠病人皮肤使用的处理头部、至少一置于处理头部内并且适于将射线投射于皮肤区域上的电磁射线源、一与该至少一个源保持热联接的热传播器、以及一与热传播器保持热联接的冷却表面。这种源可为二极管激光棒、发光二极管和灯之一。冷却表面可为一向源提供电力的导热电极的一个表面，或者可为一与源保持热联接的导热散热片的一个表面。

20 本发明的第九方面为一种用于冷却热产生装置的冷却系统，包括一与热产生装置保持热联接的冷却表面和一适于投射高压液体的喷嘴，液体形成了在冷却表面上流动的液体。这种高压液体可进行投射以便使得液体能够在位于喷嘴与冷却表面之间的整个距离上形成一液体流。

25 冷却表面可带有纹理。任选地，冷却系统可还包括一用于将液体重新引回至冷却表面上的冷却室。冷却室可包括侧壁和一盖。尽管许多实施例参照在非医学环境中执行光美容处理来进行描述，但是应当理解本发明的方面的益处既适用于非医学装置，又适用于医学装置，并且本发明还毫无限制地适用于任何一个。

30 附图说明

现在将参照附图通过实例对本发明的示例性、非限制性实施例进行描述，其中在各个图中相同的参考数字用于标示共同的元件，并且

其中：

图 1 为根据本发明的一些方面的一种光美容装置的一些基本元件的示意图；

图 2A 为根据本发明的一些方面的一种射线系统的一个实例的侧视图，用于在病人皮肤的一个区域上执行光美容程序；

图 2B 为沿着图 2A 的线 2B-2B' 剖开的病人皮肤的受到照射的区域的示意性俯视图；

图 3 为一种射线系统的一个实例的侧视图，其能够在病人皮肤的一个区域上形成两个射线区域；

图 4 为一种适用于形成多个处理岛的系统的一个实例的俯视图；

图 5 为根据本发明的方面的头部的一个实施例的示意性剖面侧视图；

图 6A 为一种使用蒸发冷却法的冷却系统的一个实施例的一个实例的剖面侧视图；

图 6B 为一种利用冷却液体的冷却系统的另一个实施例的剖面侧视图；

图 6C 为一种使用冷却液体并具有一冷却室的冷却系统的另一个实施例的示意图；

图 6D 为使用冷却液体的头部的一个实施例的剖面侧视图，其中排放孔与冷却液体进入室内所经过的口分开；

图 7 为一种具有通道的冷却系统的一个实施例的剖面侧视图；

图 8 为利用液体的蒸发冷却的头部的另一个实施例的剖面侧视图；

图 9 为根据本发明的方面使用固体相变材料的一种冷却系统的一个实施例的剖面侧视图；

图 10 为一种使用吸热化学反应来进行冷却的冷却系统的一个实施例的剖面侧视图；

图 11 为一种具有一用于冷却病人皮肤的排放孔的装置的一个实施例的剖面侧视图；

图 12A 为根据本发明的一些方面适用于光美容装置的一种单元件光学系统的一个实施例的一个实例的侧视图；

图 12B 为如图 12A 中所示的光学系统的一个实施例的一个实例的

射线轨迹;

图 13A 为根据本发明的一些方面适用于光美容装置的一种两元件柱面光学系统的一个实施例的一个实例的侧视图;

5 图 13B 为如图 13A 中所示的光学系统的一个实施例的一个实例的射线轨迹;

图 14A 为根据本发明的一些方面适用于光美容装置的一种两元件柱面光学系统的一个实施例的另一个实例的侧视图;

图 14B 为如图 14A 中所示的光学系统的一个实施例的一个实例的射线轨迹;

10 图 15A 为根据本发明的一些方面适用于光美容装置的一种两元件柱面光学系统的一个实施例的另一个实例的侧视图;

图 15B 为如图 15A 中所示的光学系统的一个实施例的一个实例的射线轨迹;

15 图 16A 为用于执行光美容程序的头部的一个示例性实施例的示意图;

图 16B 为用于执行光美容程序的头部的一个示例性实施例的示意图, 其还提供有在光美容程序进行过程中执行肌肉刺激的能力;

20 图 17A 为根据本发明的一些方面的设备的一个实施例的一个实例的示意图, 其在光学上确定着光学元件与病人皮肤表面之间的接触情况;

图 17B 为根据本发明的一些方面的设备的一个实施例的一个实例的示意图, 其在光学上确定着光学元件与病人皮肤表面之间的接触情况;

25 图 17C 为根据本发明的一些方面的设备的一个实施例的一个实例的示意图, 其在电学上确定着光学元件与病人皮肤表面之间的接触情况;

图 18A 为一种具有一运动传感器的机头的一个实施例的剖面侧视图;

30 图 18B 为一种运动传感器系统的一个实施例的一个实例的示意图;

图 19 为一种具有一光学运动传感器的设备的另一个实例的示意图;

图 20 为一种机头的一个实施例的一个实例的示意图，示出了根据本发明的一种独立式光美容装置的一些方面；

图 21 为一种用于停放独立式光美容装置的机头停放台的一个实施例的一个实例的示意图；

5 图 22 为一种具有一可分离式头部的机头的一个实施例的一个实例的示意图；

图 23 为一示意图，示出了一种适于用户更换的具有一个或多个组件的模块式机头；

10 图 24 为一示意图，示出了一种适于用户更换的具有一个或多个组件的模块式光学部件；

图 25 为一种光美容装置的一个实例的示意图，示出了本发明的一些方面；

图 26A 为一种光美容头部的一个实例的示意图，示出了针对处理皮肤的弯曲区域的本发明的方面；

15 图 26B 为用于处理弯曲表面的一种头部的双传送系统的一个实施例的示意图；

图 27 为一示意图，示出了根据本发明的机头 2700 的一些方面的一个实施例；以及

20 图 28 为根据本发明的至少一些方面的光美容装置的一个实施例的示意图。

具体实施方式

图 1 为根据本发明的一些方面的一种光美容装置的一些基本元件的示意图。区域 110 为待要执行选定的光美容处理的病人皮肤的一个区域。皮肤区域 110 具有一介于表皮层 120 与真皮层 130 之间的基底层 140。通常，光美容处理涉及处理位于表皮层 120 或真皮层 130 内的目标区域。例如，在除毛的情况中，可能需要加热毛囊 160 的球 150。另外，可能只加热球 150 的一部分，例如介于乳突与囊之间的基膜 152。

在本发明的一些实施例中，装置 100 的主子系统包括一机头 170、一基本单元 120 以及用于将机头 170 联接于基本单元 120 上的软线 126。基本单元 120 可包括一用于向控制电子电路 122 和电磁射线 (EMR) 源 125 供电的电源 124。电源 124 可以通过软线 126 联接于机头 170 上。软线 126 优选为轻型和柔性。另外，如下文中参照图 21

所述, 软线 126 可以省去而基本单元 120 可用作一位于机头 170 中的可充电式电源 (例如电池或电容器) 的充电站。在一些实施例中, 通过在机头 170 中包括一可充电式电源和一交流适配器就能够完全除去基本单元 120。

- 5 机头 170 包括一适于与病人皮肤保持接触的处理头部 180 (也简称作头部)、以及一可由操作者抓住以便在病人皮肤上沿任意方向移动头部 180 的手柄 190。例如, 头部 180 可以沿向前方向 105 在皮肤上推动, 或者沿向后方向 106 在皮肤上拉动。通常, 在给定的行程期间, 在头部 180 运动时在头部 180 和病人皮肤 110 之间仍会保持接触。
- 10 机头 170 可以通过机械驱动或者用手扫过皮肤表面的区域 110。优选在头部 180 与皮肤 110 之间保持牢固的接触以便保证良好的热和光学接触。如下文中更详细所述, 在本发明的一些实施例中, 头部 180 和/或皮肤区域 110 通过被动或主动冷却设备进行冷却以防止损坏头部并且减少损害皮肤情况 (例如伤害) 的发生。

- 15 在一个示例性实施例中, 源 125 位于机头 170 中, 例如头部 180 中。另外, 源 125 位于基本单元 120 中并且通过光学纤维 128 连接至头部 180。光学纤维 128 可以延伸通过手柄 190, 或者另外可以连接至头部 180 以便向病人皮肤输送光。

- 在一些实施例中, 控制器 122 通过线 132 接收来自头部 180 的信息, 例如关于头部 180 与皮肤 110 的接触情况、头部 180 在病人皮肤上的运动速率、和/或皮肤温度的信息。控制器 122 可以通过线 132 向头部 180 传送控制信号。线 132 可为还穿过手柄 190 连接至头部 180 上的缆的部分, 或者可以另外连接至头部。控制器 122 还可以产生输出以便控制源 125 的操作并且还可以接收来自源的信息。控制器 122
- 20 还可以控制选定的输出装置 119, 例如音频输出装置 (例如蜂鸣器)、光学输出装置、感觉输出装置 (例如振动器)、或者其它对操作者的反馈控制。根据操作者的偏好, 还可以使用其它常用的输出装置。在一些实施例中, 输出装置 119 位于机头 170 中。

- 图 2A 为根据本发明的一些方面的一种照明系统 200 的一个实例的侧视图, 用于在病人皮肤的一个区域 110 上执行光美容程序。图 2B 为沿着图 2A 的线 2B-2B' 剖开的病人皮肤的受到照射的区域 110 的示意性俯视图。在本发明的示例性实施例中, 包括一 EMR 源 204 的系
- 30

统 200 位于光美容装置的头部（例如图 1 中的头部 180）中以便使得 EMR 源位于接近皮肤表面 110 的位置。

根据待执行的处理，源 204 可以适于发射单个波长、多个波长或者一个波段。源 204 可为相干光源，例如红宝石、紫翠玉或其它固态激光器、气体激光器、二极管激光棒或其它适用的激光源。另外，源 204 可为非相干光源，例如发光二极管、弧光灯、闪光灯、荧光灯、卤素灯、卤化物灯或其它适用的灯。

由多个光学元件组成的光学系统 206 包括一用于传送 EMR 源 204 发出的射线并且与病人皮肤 110 相接触的表面 207。光学系统 206 的其它细节在下文中参照图 12-16 给出。短语“光学系统”在此用来指的是用于传送适用于执行光美容程序的任一类型光学射线的系统。

在一些实施例中，源 204 具有一沿 x 方向的延伸尺寸（例如，光源基本为线性）。本领域的普通技术人员应当理解可将多个点源组合形成一基本为线性的源。此外，较小的线性源可以组合形成单个更长连续线性源，或者具有一个或更多断点的更长线性源。例如，源 204 可为一具有 1 厘米长发射谱线和几微米谱线宽的二极管激光棒；任选地，源 204 可包括两个或三个置于一沿 x 方向的直线上的棒以形成一 2 厘米或 3 厘米长的发射谱线。

另外，线性光源可沿 y 方向彼此邻近放置以便形成一具有增大的谱线宽度的源。系统 200 可包括一个或多个另加的源 205，它们的配置与一个或更多源 204 相同或者不同。在具有两个源的实施例中，源 204 和源 205 可发射相同或不同的波长范围。

在具有多个 EMR 源 204、205 的实施例中，可能只需要启动选定的光源以便进行给定的处理。例如，在具有发射不同波长的源的实施例中，对于某些应用，例如除毛操作，可能优选只启动选定的一个或更多源，而对于某些其它应用，例如粉刺处理或皮肤恢复活力处理，优选启动选定的一个或更多其它源。尽管所讨论的源发射的是一个波长的射线，但是本发明所述领域的普通技术人员应当理解任意射线源可产生一有限范围波长的光，相应地特定的波长可为更宽范围的一部分。

射线源 204 可为一脉冲或连续波（CW）源。对于需要覆盖大面积皮肤的应用例如除毛应用，可优选 CW 二极管激光棒。在授予

Altshuler 等人的标题为“用于皮肤病学处理的方法及设备”的美国专利 U. S. Patent No. 6,273,884 B1 中, 描述了一种利用连续波 (CW) 光源来处理各种皮肤病学病症的方法, 其内容在此引入作为参考。该专利的一些方面讲述了将一 CW 光源与一接触光学输送系统结合使用, 该接触光学输送系统可用手扫描或者机械驱动掠过皮肤表面以便在目标生物结构中形成精确的温度升高 (即使用连续接触扫描 (CCS))。

大多数商用的二极管激光棒呈现的寿命大于 5000 小时, 但是根据本发明的应用可能只需要 10-100 小时的寿命。相应地, 在本发明的一些实施例中, 源 204 可以施加过载电流以便增大射线输出, 从而使得二极管激光器在较高的温度下工作, 因而造成牺牲寿命。

适用于本发明的二极管激光棒包括发射 790-980 纳米的波长或其它适当波长的二极管激光棒。适用于本发明的方面的二极管激光棒的源的实例包括 Coherent Inc. of Santa Clara, CA 或者 Spectra Physics of Mountain View, CA。源 204、205 的以上实例仅为示例, 并且应当理解本发明的方面包括使用目前已有或者尚待开发的任一适当的 EMR 源的装置和设备。

对于本发明的一些实施例, 例如需要低功率或者用于处理小面积的病人皮肤的那些实施例, 可以使用发光二极管作为光源 204、205。有宽发射波长范围的发光二极管可供使用。与以上所讨论的二极管激光器源相似, 可以在单个光学系统中使用发射不同波长的多个发光二极管。发光二极管的典型寿命范围为 50000 小时; 与激光器二极管相似, 可以通过使发光二极管过载并牺牲寿命而产生较高的光学功率。对于需要高功率密度的应用, 可以使用反射集中器 (例如抛物面反射器) 来减小在皮肤表面上的光点直径。

宽带源 (例如低功率卤素灯、弧光灯和卤化物灯) 为另一类型能够用作源 204、205 的光源。可以使用一根或更多光学纤维来为给定的应用提供感兴趣的波段。多个灯可以组合使用以便产生高功率, 并且与发光二极管的情况相似, 可以使用集中器来减小在皮肤表面上的光点直径。在一些实施例中, 几种不同类型的光源可以并入一光美容装置 (例如图 1 的装置 100) 中。

在系统 200 的一些实施例中, 一束分离器 230 将来自源 204 的

射线分离形成 EMR 的第一部分和 EMR 的第二部分。第一部分和第二部分可以各自通过过滤器 240 和 242 进行过滤。在过滤之后，这两个部分可以具有相同或者不同的波长范围。第一和第二部分的功能可以相同或者不同。例如，EMR 的第二部分的功能可为对病人皮肤进行预热，以便为利用 EMR 的第一部分进行处理作准备。另外，EMR 的第一部分和 EMR 的第二部分可以都提供处理。

参看图 2B，在一些实施例中，光学系统 206（在图 2A 中可以看到）适于在病人皮肤 110 上形成一沿着第一轴线 211 的射线第一区域 210。射线第一区域 210 由来自源 204（在图 2A 中可以看到）的电磁射线的至少第一部分形成。在一些实施例中，可在病人皮肤 110 上形成一沿着第二轴线 221 的射线第二区域 220。射线第二区域 220 可由来自射线源 204 的电磁射线的第二部分形成；另外，射线第二区域 220 可以由来自第二射线源 205（在图 2A 中可以看到）的光形成。

在本发明的一些方面中，第一轴线 211 和第二轴线 221 平行；然而在其它实施例中，轴线 211、221 并不平行。系统 206 可适于使第一区域 210 与第二区域 220 形成一选定距离，或者可适于使射线的第一部分覆盖着射线的第二部分的至少一部分。任选地，系统 206 适于基本上以线的形式形成（聚焦或校准成）第一部分和第二部分。光学系统 200 可以适于在皮肤表面上产生一条或更多条光线，其各具有 1-300 毫米的长度和 0.1-10 毫米的宽度。光束的散光可处于 0.01-0.5 的范围内。词“散光”在此的定义指的是光束宽度与光束长度的比率。同样，任选地，系统 206 可适于在病人皮肤 110 上形成一个或更多沿着其它轴线（未示出）的射线的其它区域，射线的其它区域由来自射线源 204 或 205 的电磁射线的相应其它部分，或者来自一个或更多其它射线源的射线形成。

图 3 为一种用于执行光美容程序的照明系统 300 的另一个实例的侧视图，其能够在病人皮肤 110 的一个区域上形成两个射线区域 311、316。在系统 300 中，两个光学系统 310、315，而不是单个光学系统 206（图 2），各产生一相应的射线区域 311、316（例如射线区域 210、220）。用于产生线的射线可来自两个源 304、305 或者以上参照图 2 所述的单个分开的源。

图 4 为一种适用于形成多个处理岛的照明系统 400 的一个实例的

俯视图。系统 400 包括多个源 410 (例如一发射照明线或圆斑点的常规型激光器二极管), 其各具有一相应的光学系统 415 以便将光引至皮肤区域。所示的系统可用于在皮肤区域内形成一具有多个处理岛的焦斑的正方形(或任意形状)矩阵。此处所用的词“岛”的定义指的是与进行指定处理的其它区域分离开的进行指定处理的一个区域, 因此使得位于两个或更多区域之间的区域所接收的射线量低于为实现指定处理所需的量。对于照明岛, 在由 Anderson 于 2001 年 12 月 27 日提交的标题为“用于 EMR 处理的方法及设备”的美国临时专利申请 U. S. Provisional Patent Application 10/033,302 中进行了更详细的讨论, 其内容在此引入作为参考。

对于利用高功率源的根据本发明的光美容装置的实施例, 为了防止对消费者造成伤害或其它损害, 来自源的废热的管理非常重要。例如, 在机头中包括二极管激光棒的光美容装置的情况中, 可能有高达 60% 的电能以非光学废热的形式耗散。除了要除热以防伤害之外, 除热对于防止源过热和源寿命缩短也很重要。

图 5 为根据本发明的方面的头部 500 的一个实施例的示意性剖面侧视图。头部 500 包括一照明系统, 其包括一 EMR 源(例如二极管激光棒 510)和一光学系统 520。头部 500 可位于一壳体中以便保护光学组件并且保护光美容装置的操作者; 壳体在此省去以免造成混乱。在图 5 中, 二极管激光棒 510 担当电磁射线源(例如图 2 中的源 204)并且可以用来形成一个或更多射线区域(例如图 2 中的 210、220)。二极管激光棒 510 位于正电极 515 与负电极 516 之间。电极 515、516 为二极管激光棒 510 提供电功率, 并且可由具有良好导电性的任意适用材料制成。在一些实施例中, 电极 515、516 与二极管激光棒 510 保持热接触, 并且具有良好的导热性以便将废热从二极管激光棒 510 处送离。例如, 电极 515 和 516 可由铝或铜制成。

任选地, 来自二极管激光棒 510 的废热可通过电极 515 和 516 传送至散热片 530。散热片 530 可由具有良好导热性的任意材料制成以便将废热从二极管棒 510 处送离。例如, 散热片可由铝或铜制成。散热片 530 可通过任意适当的已知冷却方法包括空气流来进行冷却。任选地, 冷却作用可通过向散热片 530 添加翼片(未示出)而得以增强。另外, 散热片 530 也可以通过一种或更多以下参照图 6-11 所讨论的除

热方法来冷却。同样任选地，一热传播器 522 可位于电极 515、516 与散热片 530 之间。热传播器 522 与电极 515、516 和散热片 530 保持热联接。热传播器 522 可由任意具有良好导热性的适用材料制成；优选地，热传播器 522 为电绝缘。金刚石和碳纤维为适用作热传播器的材料的两个实例。

在一些实施例中，电极 515、516 适于为散热片以便将废热从二极管激光棒 510 处传离。相应地，散热片 530 和热传播器 522 可以省去。在这些实施例中，电极 515 和 516 可以由任意具有良好导热和导电性的材料制成。任选地，可以使用一个或更多热传感器 524（例如热电偶、电热调节器）来监控指示病人皮肤的温度（例如在光学系统 520 与电极 516 的界面处的温度）以用于下文所述的冷却系统中。

二极管激光棒 510 可以利用能够在棒 510 与电极 515、516 之间保持良好电接触的任意方法而固定于电极 515 和 516 上。在需要传送废热的实施例中，可以使用能够实现良好热和电接触的任意适用方法。在一个实施例中，二极管激光棒 510 夹紧于两个电极 515 和 516 之间。可以利用弹簧或其它适用的装置来将二极管激光棒 510 牢固夹紧于电极 515、516 之间。在另一个实施例中，二极管激光棒 510 也可利用导热/电环氧树脂而胶合就位。在另一个实施例中，二极管激光棒 510 利用低温焊料（铟或金/锡焊料，等等）而焊接就位。可以利用置于二极管激光棒 510 与电极 515 和 516 之间的铟的预型，并利用模结合器来加热、压缩、然后冷却焊料和二极管棒而实现自动焊接。任选地，一由具有高导热性以及低导电性的材料如 BeO 制成的垫片 525 可以包括在内以便在电极 515 和 516 之间提供电绝缘。

根据本发明的一些方面，光学系统 520 将来自二极管激光棒 510 的光联至病人皮肤。光学系统 520 可以通过气隙 511 与二极管激光棒 510 分离。在下文中参照图 12-15 对示例光学系统 520 进行了更详细的描述。在根据本发明的一些方面的实施例中，光学系统 520 适于与病人皮肤的一个区域相接触，并且对光学表面 521 进行冷却以提供对病人皮肤的冷却作用。

在一些实施例中，二极管激光棒 510 和光学系统 520 的冷却利用单个冷却系统来实现。例如，电极 515、516 可沿尺寸 A 与光学系统 520 保持热联接；相应地，二极管激光棒 510 和光学系统 520 都可以

通过直接冷却电极 515、516 或者通过冷却与电极 515、516 保持热联接的散热片 530 而进行冷却。尺寸 A 通常大约介于 1 至 10 毫米之间。关于光源和光学系统的同时冷却的其它细节在 1999 年 12 月 28 日提交的序号为 09/473,910 的美国专利申请中给出，其内容在此引入作为参考。

皮肤的接触冷却可用来在向皮肤输送例如具有能够有效吸收黑色素的波长的高流量射线的过程中保护病人的表皮。在头部 500 的一些实施例中，光学系统 520 包括一蓝宝石元件，由于其具有良好的光传输能力和导热性而适于与病人皮肤相接触。如上所述，光学系统 520 可在处理过程中进行冷却以便从蓝宝石元件除去热量。任选地，在利用光美容装置进行处理之前，可将一种在工作波长下透明的洗液涂敷于皮肤上。优选地，该位置能够导热以便增强通过光学表面 521 从皮肤上除热的能力。优选地，这种洗液还便于光学系统 520 在皮肤表面上的滑动，并且具有适合于接触表面 520 与皮肤 110 之间相匹配的折射率以便有效地提供光学联接至皮肤中的射线。

通过选择一种具有响应于 EMR 源（例如激光二极管 510）的辐射而改变的光学属性（例如颜色或反射系数）的洗液，洗液还可用于显示哪块皮肤区域已进行过处理。例如，如果洗液开始时呈现一种给定的颜色，在经过照射后其将变透明（或呈一种不同的颜色）。区分已处理区域和未处理区域的能力对于需要在较大表面积上执行的处理例如除毛而言尤其重要。

图 5 还示出了一种用于通过散热片 530 冷却二极管棒 510 和光学系统 520 的系统的实施例。在图 5 中，吸热液体流过与散热片 530 保持热联接的导热管道 540。例如，在一个实施例中，水用作这种液体。任选地，水可以通过连接一个冷水源，例如自来水而提供；参看图 1，水可以通过具有适当的管路的手柄 190 提供。另外，一具有热交换器（未示出）的闭路冷却回路用于除去液体的热；热交换器可以位于手柄 190 或基本单元 120 中。

再次参看图 5，管道 540 覆盖着一个或更多表面，例如散热片 530 的表面 542 的至少一部分。单个平管道可以覆盖着散热片 530 的一个或更多表面的全部。另外，可以使用多个管道，而每个管道覆盖着散热片 530 的一个表面的一部分。另外，一个或更多管道 540 可以覆盖

着电极 515、516 的至少一部分。由于冷却作用可以应用于散热片 530 或者直接应用于电极 515、516 上,因而散热片的表面(例如表面 542)、电极的表面或者其它有待除热的适当表面在后文中都将称作“冷却表面”。尽管所示的冷却表面为一外表面,但是应当理解,冷却表面可
5 可为内表面,例如通过散热片或电极暴露于管道的表面。

图 6A 为一种使用蒸发冷却法的冷却系统的一个实施例 600 的一个实例的剖面侧视图。在图 6 中,相变液体从一个或更多喷口 610 和 620 喷至冷却表面 623 上。这种液体可为任意适用的蒸发液体,因而液体响应于从冷却表面上吸热而蒸发。在一些实施例中,这种液体为
10 一低温沸点液体,其被引至散热片上以便使得液体响应于从冷却表面 623 上吸热而沸腾。在一些实施例中,这种液体为四氟代乙烷(沸点-26℃)、二氧化碳(沸点-78℃),但也可使用任意其它适用的液体(例如氟利昂或液氮)。在一些实施例中,液体通过喷口 610 和 620 而喷成雾状。

任选地,液体可以容放于一位于基本单元或手柄中的容器 625 中。优选地,容器 625 便于用户够到以便可由用户更换。管道 626 用于将液体送至喷口 610 和 620。冷却液流量通过阀 627 进行调节,其利用关于系统(例如图 5 的系统 500)中存在的热量的信息手动或电动控制。例如,可利用一传感器(例如图 5 中的传感器 524)来控制阀 627
20 中的反馈控制螺线管。任选地,每个喷口 610 和 620 可为一种组合阀喷口,从而无需单独的阀 627。

任选地,进行蒸发的冷却表面 623 可以具有纹理以便增大可以蒸发液体的表面积。尽管所示的蒸发表面为三角形纹理 615,但是可以采用任意适用于增大表面积的形状。所示的三角形纹理 615 可为一
25 线性凹槽型式、一同心圆形凹槽型式的截面或任意其它适当的凹槽型式的一部分。其它纹理包括多个凸起(例如,从冷却表面凸出的半球、圆柱或锥)。任选地,可使用一环 630 来环绕着喷口 610、620 和散热片 530 以便容放喷雾。

还可使用一相变液体来冷却用于对光美容装置供电和/或进行控制的电子电路 644。特别地,用来控制光美容装置的功率的功率场效应晶体管(FETs)产生大量的热。按照常规,利用较大的散热片和一风
30 扇来冷却功率场效应管以便除热。这种系统容易会又大又重。根据本

发明的冷却系统提供了另一种替代冷却方法。

任选地，提供用于除去由 EMR 源所产生的热的液体的相变液体管道 626 的一部分可以适于将一部分相变液体引至喷口 640。喷口 640 将一部分相变液体引至冷却表面（例如散热片的表面 642）上。可使用一热传感器 646（例如电热调节器）来控制喷至冷却表面上的液体量，例如通过控制阀 650 来控制。

图 6B 为一种利用流过的冷却液体 605 的用于头部中的冷却系统的另一个实施例 650 的示意图。在图 6B 中，高压液体保持于一容器 655 中（例如 1 至 5 个大气压的四氟代乙烷）并且通过喷嘴 660 喷至冷却表面 665 上。从喷嘴 660 喷出的液体 607 可为液体小滴或流的形式。在一些实施例中，液体以流的形式喷出以克服小滴的空气动力学属性（即高阻力），从而改进冷却系统 650 的除热属性。如上所述，冷却表面 665 可为任意热的良导体的材料（例如铜或银）。优选地，所选定的冷却表面 665 具有足够大的尺寸以便使得液体 655 从表面 665 上蒸发而非从所述表面上滴完。

从喷嘴 660 喷出的液体 607 被喷至冷却表面 665 上以便在冷却表面 665 上形成流动液体 605。可以选定喷嘴 660 和冷却表面 665 以便使得从喷嘴 660 喷出的液体 607 为一沿介于喷嘴 660 与冷却表面之间的整个距离上的液体流，并且在撞击表面 665 时在冷却表面 665 上形成流动液体。另外，可以选定喷嘴 660 和冷却表面 665 以便使得从喷嘴 660 喷出的液体 607 可在聚集于冷却表面 665 上形成流动液体之前在喷嘴 660 与冷却表面 665 之间形成喷射的小滴。由于从喷嘴 660 喷出的液体处于高压之下，因而在冷却表面 665 上的流动液体以较高的速度 V 流过冷却表面 665。

利用在冷却表面 665 上形成流动的液体 605，与在冷却表面 665 上形成小滴（即非流动的液体）的常规型冷却系统相比，就能增大从表面 665 上除热的能力。例如，除热能力改进的原因在于小滴（形成于常规型系统中）不能形成足够大的数量和密度以实现和保持选定的除热量。

图 6C 为一种使用冷却液体 655 并具有一冷却室 684 的用于头部中的冷却系统的另一个实施例 670 的示意图。头部 670 具有侧壁 675 和顶盖 680，顶盖 680 具有一用于使来自喷嘴 660 的液体 655 进入的

口 682。侧壁 675 和顶盖 680 形成室 684。口 682 也可用作蒸发的冷却液体的排放孔。如箭头 686 所示，侧壁 675 和顶盖 680 将来自顶盖 680 的液体 655 重引回至冷却表面 665 上。所选定的侧壁 675 优选地与冷却表面 665 保持热联接以便使得与侧壁 675 相接触的液体可以除去冷却表面 665 上的热。任选地，侧壁 675 可以与冷却表面 665 形成一体以便使得与侧壁 675 相接触的液体能够除热。在一些实施例中，可能优选顶盖 680 具有较差的导热性和较差的对冷却液体的润湿特性以便提高冷却液体到达冷却表面 665 的可能性。例如，在一些实施例中，顶盖 680 由聚合物或有机玻璃制成。尽管所示的室 684 的侧壁和顶盖之间形成一个角度，但是所形成的室也可以具有连续的曲率。

由于口 682 担当蒸发液体的排放孔的作用，因而口 682 的面积 S 决定着室 684 内所保持的压力。在一些实施例中，所选定的口 682 具有足够大的面积 S 以便防止回压降低液体喷至冷却表面 665 上的速度；然而，所选定的口 682 可以足够小以便使得顶盖 680 能够将大部分的液体重引回至冷却表面 665 上，并且保持着室 684 内的压力以便防止液体蒸发过快。例如，口面积 S 可约为喷嘴 660 的面积的一百倍至两百倍。在一些实施例中，对于低于或等于大气压的压力，所选定的冷却液体为一种沸腾温度（即蒸发温度）低于 -26 摄氏度的液体。

图 6D 为使用冷却液体的激光头的一个实施例 690 的剖面侧视图，其中排放孔 692 与冷却液体进入室 696 内所经过的口 694 分开。室 696 由冷却表面 688、侧壁 693 和顶盖 695 所界定。冷却表面 688 通过联接板与源 525 和光学系统 520 保持热联接（在下文中进行更详细描述）。来自喷嘴 698 的冷却液体喷至带纹理的冷却表面 688 上。未与冷却表面 688 相接触的一部分冷却液体由侧壁 693 和顶盖 695 直接重引，如箭头 686 所示。

任选地，所选定的顶盖 695 具有一谐振频率以便增强将液体重引至冷却表面 688 的能力。同样，任选地，一用于降低液体动能的装置（例如螺旋桨，未示出）可置于喷嘴 698 与冷却表面 688 之间以便冷却液体。

图 7 为一种用于接触皮肤表面 110 的头部的一个实施例 700 的剖面侧视图。头部 700 具有位于电极 515、516 中的通道 730 和 731。蒸发冷却可沿电极 515、516 的底面以及沿通道 730、731 的表面进行，

从而增加头部 700 的冷却表面积。优选地，通道 730 和 731 的位置靠近二极管激光棒 510。在一个实施例中，通道 730、731 沿着二极管激光棒 510 的长度（即沿着 x 方向）定位。在一些实施例中，通道 730、731 的位置靠近喷口 610 以便接收喷射液体。通道 730 和 731 可具有
5 矩形横截面或任意适用于改进冷却作用的其它形状。例如，开口 740、742 可以张开接收来自喷口 610 的喷射液体。作为沿二极管棒 510 的长度延伸的单根通道的一种替代方案，一组通道可以沿着二极管激光棒的长度置于二极管激光棒 510 的一侧或两侧。

图 8 为冷却系统的另一个实施例 800 的剖面侧视图。在图 8 中，
10 利用液体来从冷却表面 823 上除热但是液体并非以喷射形式使用。在所示的示例性实施例中，液体从容器 825 流出至多个位于冷却表面 823 内的通道 832 中。这些通道 832 中每一个的长度沿源 510 的长度方向延伸。液体与冷却表面 823 形成热接触或物理接触。

任选地，液体可为一种响应于从冷却表面 823 上吸热而蒸发的低
15 沸点液体。可利用阀 833 来控制液体蒸发情况；当需要特别有效的冷却时，阀 833 打开并且对液体施加低于平衡的压力以利于蒸发。压降引起液体沸腾，而这就会从冷却表面 823 上除热。尽管所示的通道 832 沿平行于光源 510 的长度的方向延伸，并且所示的通道具有矩形横截面，但是也可以采用沿一个或更多不同方向对齐的其它形状的通道
20 832，其都在本发明的本方面的范围之内。可通过一热传感器（例如图 5 中的传感器 524）而得到一反馈信号来控制位于控制阀 833 中的螺线管。

图 9 为一种用于接触皮肤表面 110 的头部的另一个示例性实施例 900 的剖面侧视图。头部 900 具有一冷却系统，该冷却系统具有一与
25 固体块（也称作相变固体）形成物理接触的冷却表面 923。固体块 834 的至少一部分响应于从冷却表面 923 上吸热而发生相变。相变可为从固体变为液体，或者从固体变为气体。在一些实施例中，固体的熔化温度介于大约 -10C 至 +30C 之间；然而，在一些应用中，可以利用在这个范围之外，特别是在这个范围之下发生相变的材料。

30 在一些实施例中，固体块方便地位于装置机头（例如图 1 中的机头 170）内以便可由用户更换。在一些实施例中，固体块容放于一绝缘套筒中以避免接触用户的手，并且/或者将暴露于室温下的熔化过程

减至最小。在所示的实施例中，温度控制可通过使用用于使固体块与冷却表面 923 形成和脱离接触的一手动或电动控制的螺线管或一弹簧 835 来实现。

5 在相变冷却系统的一个实施例中，相变固体为冰。在这个实施例中，用户可以在其冷冻器中存放一块或更多块冻结冰块。当用户想要操作光美容装置时，可以将一块冻结冰块插于装置中。在另一个实施例中，也可以使用熔点比水低得多的干冰来实现更大的冷却能力。应当理解，冰块可以包含水或者带有一种或更多添加剂的水以便对用户皮肤进行处理。

10 在一些实施例中，可以使用市场上可买到的有机化合物（例如石蜡基的材料、脂肪酸、交联型聚乙烯）作为相变固体。适用的石蜡材料的实例包括由 Rubitherm GmbH 所生产的 RT25。RT25 的熔点为 27.7℃。在其它实施例中，可利用熔点处于 20-35℃ 范围内的油脂作为相变固体。在另一个实施例中，利用适于具有在 15 - >50℃ 范围内的
15 熔点的镓或镓合金（例如 Ga/In、Ga/In/Sn 或 Ga/In/Sn/Zn）来作为固体块。在 Ga/In 合金中，Ga（40.6 W/m*K）和 In（81.6 W/m*K）的较高导热性将有助于将废热传遍合金体积。可以使用一次性相变冷却器筒来容纳相变固体；例如，相变固体可以使用一次然后丢弃或者可以可再充装（即重凝固一次或更多次）。

20 图 10 为头部的一个实施例 1000，其具有一使用吸热化学反应来进行冷却的冷却系统。适当的反应实例为将硝酸铵（ NH_4NO_3 ）或氯化铵（ NH_4Cl ）放入水中所产生的吸热反应。例如，如果 200 毫升的水与 200 克的硝酸铵混合，可以达到大约为 -5℃ 的温度，从而容许吸热。

25 在图 10 中，吸热反应包含于一反应室 1050 内，而反应室与冷却表面 1023 保持热联接。在一些实施例中，反应室 1050 可通过具有良好导热性的材料与冷却表面 1023 相联接。在一些实施例中，这种机构包括一薄膜 1051，其将第一室的水与另一室的氯化铵隔开。在一些实施例中，膜 1051 可以破裂以便启动化学反应而反应室可为一次性的容器。
30 例如，用户可以对柔性塑料反应室施加力以便使膜破裂从而在开启装置之前产生一容器的冷却液体。另外，膜也可以除去或者根据已知的工具进行操纵以便容许第一室和第二室的内含物相互反应。

图 11 为一种具有一管道 1110 和一排放孔 1120 的装置的一个实施例 1100 的剖面侧视图。在图 11 中，进入排放孔 1120 的液体或气体被引至皮肤区域 1130，以便在处理过程中对皮肤区域 1130 进行预冷却或后冷却。例如，喷至冷却表面 530 上的相同冷却液体的一部分或者由液体蒸发所产生的气体可以进入管道 1110 并且通过孔 1120 喷至皮肤上。液体部分可为浓缩的蒸汽或者仅为过量的液体。如上所述，如果使用自来水来进行冷却（或者如参照图 9 所述的冰相变冷却器），就可以在水用于对冷却表面 530 进行冷却之后使一部分水转向。在一些实施例中，来自相变冷却系统产生的气体的压力可用来将洗液送至病人皮肤上。尽管所示的实施例示出的是在液体用于冷却表面 530 之后使一部分冷却液体转向，但在一些实施例中，可以直接将一部分冷却液体喷至皮肤上而不用于对冷却表面 530 进行冷却。

任选地，在喷至皮肤上之前，可通过管道 1112 将一种或更多种添加剂加入液体中（例如，用于形成一冷却洗液）。添加剂可以存储于机头或基本单元中的筒（未示出）内。在一些实施例中，为了实现“淋浴效果”，从散热片放出的水可以全部排至皮肤上。作为使用蒸发液体的一种替代方案，可以将一替代气体、液体或洗液源（即独立于冷却系统）存储于机头或基本单元中的筒内并且当机头移过皮肤表面上时放出。

为了避免困惑，将参照单个电磁射线源对用于本发明的方面的光学系统的以下示例性实施例进行描述；然而，如上所述，可以使用一个或更多源来形成一个或更多射线区域。在以下所述的示例性光学系统中，每个具有光学功率的表面沿第一轴线（例如 y 轴）具有光学功率而沿垂直于第一轴线的轴线（即 x 轴）则具有零光学功率。就是说，透镜为圆柱形。尽管以下所讨论的实施例具有平面或圆柱形曲率，但是其它折射或衍射光学设计也在本发明的范围之内。

图 12A 为根据本发明的一些方面适用于光美容装置的一种单元件光学系统的一个实施例 1200 的一个实例的侧视图。光学系统 1200 包括一用于从电磁射线源 1220（例如激光二极管棒）向病人皮肤 110 传送光的元件 1210。元件 1210 具有一输入表面 1211 和一适于与病人皮肤表面接触的输出表面 1212。

源 1220 靠近耦联于元件 1210 的输入表面 1211 上（例如隔开 1

毫米); 这种靠近耦联使得沿着激光二极管源的高度发散快轴的光的一大部分被传送至病人皮肤。在一些实施例中, 输入表面 1211 具有一抗反射 (AR) 涂层。

5 如上所述, 元件 1210 由在工作波长下基本透明的材料制成, 并且优选地由能够导热以便从经过处理的皮肤表面上除热的材料 (例如蓝宝石) 制成。在一些实施例中, 元件 1210 的侧向侧 1213 上涂有在工作波长下具有反射能力的材料 (例如铜、银或金)。另外, 位于源 1220 与输入表面 1211 之间的空间 1221 可以用反射材料包围住以便增加入射于表面 1211 上的光的强度。

10 在一个实施例中, 光学元件 1210 为一蓝宝石板 (即, 表面 1211 和 1212 为平面, 并且不具有光学功率)。在光学系统的另一个实施例 1200 中, 光学表面 1212 具有一圆柱形曲率 (如图 12 中所示) 并且进行选择以便会聚入射于表面 1212 上的光。例如, 在一个实施例中, 表面 1212 的曲率半径大约为 3 毫米。这种系统可以用于处理需要高
15 处理能量密度的皮肤结构。例如, 图 13 的透镜系统可以用于对准毛囊的干细胞、皮脂腺、下漏斗管、血管组织、纹身或胶原蛋白。

在一些实施例中, 侧向表面 1213 的长度 L 大约在 5-50 毫米的范围之内, 并且对横截面宽度 (沿 x 方向测量) 和高度 (沿 y 方向测量) 进行选定以便聚集来自源 1220 的光。例如, 对于包括两个靠近
20 联接于元件 1210 上的 1 厘米二极管激光棒的源, 横截面宽度选定为 2 厘米, 而横截面高度为 2 厘米。

如图所示, 光学元件 1210 将来自源 1220 的一部分光直接传送至表面 1212 上而不在侧向表面 1213 上反射 (例如示例性的射线 1230), 并且来自源 1220 的一部分光在到达表面 1212 之前由侧向表面 1213
25 进行反射 (例如, 示例性的射线 1232)。利用全内反射将来自源的一部分光引至表面上的元件, 例如元件 1210, 在此也称作“波导”。

任选地, 可添加一尖端反射器 1222 以便将散射于皮肤之外的光重新引回至皮肤上 (称作光子再循环)。对于近红外的波长, 介于 40 % 至 80 % 之间的入射于皮肤表面上的光被散射于皮肤之外; 作为一名
30 普通技术人员应当理解散射量部分地取决于皮肤着色情况。通过利用尖端反射器 1222 将散射于皮肤之外的光重新引回至皮肤上, 由系统 1200 所提供的有效能量密度可以增加两倍以上。在一个实施例中, 尖

端反射器 1222 从元件 1210 的上侧向表面和下侧向表面上延伸总长为 3 毫米。在一些实施例中，尖端反射器 1222 具有一铜、金或银涂层以便将光反射回至皮肤上。

5 反射涂层可以应用于暴露于来自皮肤的反射/散射光之下的装置的任意非传送表面上。本发明所述领域的普通技术人员应当理解，这些表面的位置和功效取决于光源的所选聚焦几何和位置。光子再循环在 Altshuler 等人于 2000 年 8 月 9 日提交的标题为“用于皮肤病学处理的头部”的美国申请 09/634,981 和 1999 年 3 月 12 日提交的序号为 09/268,433 的申请中进行了进一步的讨论，两者的内容在此引入作为
10 参考。图 12B 为如图 12A 中所示具有一源 1220 和一元件 1210 的光学系统的一个实施例 1200 的一个实例的射线轨迹。

图 13 为根据本发明的一些方面适用于光美容装置的一种两元件柱面光学系统的一个实施例 1300 的一个实例的侧视图，其中一准直器 1310 与元件 1210 协同使用。在图 13 中，一快轴准直器 1310 非常
15 靠近地耦联于光源 1220（例如 0.09 毫米）。在一个实施例中，准直器 1310 具有一 1.5 毫米的长度、一平面输入表面 1311 以及一输出表面 1312，其中输出表面 1312 具有一曲率以便对准直器 1310 的输出进行准直。元件 1210 位于与输出表面 1312 距离 0.1 毫米的位置。准直器 1310 产生的射线光束基本上在输出表面 1312 处沿 y 方向准直。例如，
20 准直器 1310 可为由德国的 Limo GmbH of Dortmund 公司生产的号为 S-TIH53 的透镜模块。

准直束投射于光学元件 1210 的输入表面 1211 上。如上所述，元件 1210 可为一板或者可稍微发散（例如输出表面 1212 可以具有等于 3 毫米的曲率半径）以便补偿在皮肤中的散射。这种系统可用于处理
25 需要高处理能量密度的皮肤结构。例如，图 13 的透镜系统可用于对准毛囊的干细胞、皮脂腺、下漏斗管、血管、纹身或胶原蛋白。图 13B 为如图 13A 中所示的这种具有一源 1220 和一准直器 1310 和一元件 1210 的光学系统的一个实施例 1300 的一个实例的射线轨迹。

图 14A 为根据本发明的一些方面适用于光美容装置的一种两元件
30 柱面光学系统的一个实施例 1400 的另一个实例的侧视图。在光学系统 1400 中，图 13 的快轴准直器 1310 和与准直器 1310 的表面 1312 距离 0.1 毫米的元件 1420 协同使用以便从源 1220 投射光线。元件 1420

具有一曲率为 1 毫米的输入表面 1421、一平面输出表面 1422 以及 1 毫米的长度。系统 1400 在距离表面 1422 大约 1 毫米处聚焦光线（即，对于表面 1422 适于与病人皮肤相接触的实施例而言，位于皮肤表面 1 毫米之下）。在一个实施例中，元件 1310 和 1420 的高度选定为 1.5 毫米。在一些实施例中，透镜 1420 由蓝宝石制成。这种系统可以用于处理需要高处理能量密度的浅层皮肤结构。例如，图 14 的透镜系统可用于对准牛皮癣、皮脂腺、毛干或者毛干细胞。图 14B 为如图 14A 中所示的这种具有一源 1220 和一准直器 1310 和一元件 1420 的光学系统的一个实施例 1400 的一个实例的射线轨迹。

图 15A 为根据本发明的一些方面适用于光美容装置的一种两元件柱面光学系统的一个实施例 1500 的另一个实例的侧视图。图 15 所示的光学系统 1500 能够用于例如聚焦比图 14 中的光学系统 1400 更深的二极管光。例如，光学系统 1500 可以将二极管光聚焦于皮肤表面下方大约 2 毫米（即与表面 1522 距离 2 毫米）以便对准皮肤中的深层结构（例如毛球、深层血管、皮下脂肪）。

系统 1500 为一种用于投射来自源 1220 的光的两元件对称透镜系统。第一元件 1510 位于与源 1220 距离大约 1.4 毫米的位置处并且具有一平直的输入表面 1511 和一曲率为 2.5 毫米的输出表面 1512；相应地，透镜 1510 拟准直来自光源 1522 的光。第二透镜 1520 具有一曲率为 2.5 毫米的输入表面 1521 和一平直输出表面 1522；相应地，透镜 1522 将经过拟准直的光聚焦于皮肤表面以下 2 毫米。在所示的实施例中，对光学系统中的偏差进行平衡以便在输出表面 1522 处得到基本均匀（即“平顶”）的空间光强分布图。平顶强度分布图基本上由在垂直于圆柱形表面 1522 的平面中的球面像差确定。在一些实施例中，透镜 1510 和 1520 由蓝宝石制成。图 15B 为如图 15A 中所示的这种光学系统的一个实施例 1500 的一个实例的射线轨迹，其具有一源 1220 和一元件 1510 及一元件 1520。

图 16A 为用于执行光美容程序的头部的一个示例性实施例 1600 的示意图。所示的头部 1600 不带壳体以便于描述。如上所述，头部 1600 将沿病人皮肤的一个区域运动，通常沿方向 1602 或方向 1604 运动。

头部 1600 包括一用于传送来自一 EMR 源 1630 的光的光学系统 206。电极 1620 启动 EMR 源 1630。电绝缘器 1650 可位于电极 1620

之间以防止电极 1620 之间发生电接触。电极 1620 可以逐渐变细以便减小与病人皮肤的接触区域。

图 16B 为用于执行光美容程序的头部的一个示例性实施例 1650 的示意图，其还提供有在光美容程序进行过程中执行肌肉刺激的能力。电肌肉刺激为一种众所周知的物理治疗程序，其可以增强一些光美容程序的功效。例如，电肌肉刺激可以用来提高皱纹处理或脂肪团处理的功效。

在一个实施例中，用于输送电刺激的两个电极 1610 位于光学系统 206 的相对侧，头部 1600 的设计成在光美容处理过程中（即，在通过系统 206 输送 EMR 的过程中）与病人皮肤相接触的一部分上。一个电极 1610 在光学系统 206 之前与病人皮肤的一个区域相接触，而另一个电极 1610 在光学系统 206 之后与病人皮肤的一个区域相接触。

可以使用一种导热的电绝缘器 1615（例如由 BeO 或金刚石或其它适用的材料制成）来防止在提供电刺激的电极 1610 之间，以及在启动 EMR 源 1630 的电极 1620 之间发生电接触。一电绝缘器 1650 可位于电极 1620 之间以防止电极 1620 之间发生电接触。

通过在机头扫过皮肤表面时利用电极 1610 来向病人皮肤施加一恒定（或脉冲）电流，就可以同时实现肌肉刺激和电磁处理。在一些实施例中，电极可以提供通过皮肤的射频（RF）电流。另外，电极 1610 可以提供一直流电流或者一微波场。在一些实施例中，皮肤可以利用 RF 电流或微波场扫过以便有选择地加热待利用 EMR 射线处理的皮肤部分。预热皮肤可以使得 EMR 源 1630 的功率能够降低。

图 17A 为根据本发明的一些方面的设备的一个实施例的一个实例的示意图，其在光学上确定着光学元件 1704（例如图 12 的元件 1210）与病人皮肤表面 1701 之间的接触情况。为了提供眼睛安全性，在光美容装置的一些实施例中，使用一接触传感器来使得电磁处理源（例如图 5 的源 510）能够只在装置与病人皮肤相接触时才会启动。

在图 17A 中，一照明源 1702（例如二极管激光器或发光二极管，与处理源分开）安装于距离元件 1704 几毫米处（例如 5 毫米），并且指向皮肤表面 1701。任选地，照明源 1702 可以安装成通过元件 1704 将光引至皮肤表面 1701 上。源 1702 可以发射波长与处理源 510 相同

的射线，但是优选地发射波长与处理源 510 不同的射线。设置一检测器 1712 以便检测从皮肤表面 1701 反射或散射的来自照明源的光。任选地，可以增加一过滤器 1708 以便有选择地传送来自源 1702 的光，并消除与处理源 510 相对应的光波长以及任何其它外来的光波长。

5 在很少或者没有皮肤接触的情况下，将有较大量的来自源 1702 的射线光通过光学系统 1704 从皮肤表面 1701 反射或散射至检测器 1712 上。如图 17B 中所示，当元件 1740 与皮肤表面 1701 形成良好接触时，皮肤中的散射和吸收将会减弱来自照明源 1702 的光，因而将会有较少量的射线到达检测器 1712。因此，通过使用电子装置（例
10 如一比较器）来测量检测器 1712 的输出，并选择一适当的阈值，就能使处理源适于只有在检测器 1712 的输出位于阈值之下时才会启动。任选地，源 1702 和/或检测器 1712 可以位于基本单元中并且可以使用一个或更多光学纤维来将光从机头与源或检测器相联接。

在另一个实施例中，检测器 1712 检测来自处理源的光以便确定
15 元件 1740 与皮肤表面 1701 之间的接触情况。在这种系统中，来自源 510 的光通过元件 1704 由皮肤表面 1701 散射和反射至检测器 1712。一射线过滤器 1708 可以有选择地将这种经过散射和反射的射线传送至检测器 1712。在这个实施例中，处理源 510 保持在低功率眼睛安全模式下，直到与皮肤表面 1701 形成牢固接触为止。当在皮肤表面 1701
20 与元件 1704 之间没有或者很少接触时，检测器 1712 的输出就比较低。然而，当元件 1704 与皮肤表面 1701 形成良好接触时，检测器 1701 的输出就比较高。因此，处理源 510 将会适于只有在检测器 1712 的输出处于阈值水平之上时才会发射。

另外，也可以使用一种为光学计算机系统鼠标的标准光学接触检测器来代替源 1702 和检测器 1712，例如使用由 Logitech of Fremont,
25 CA 公司生产的 CordLess Mouseman™ 中的光学接触系统。

作为用于测定接触情况的光学方法的一种替代方案，可以使用电方法来检测元件 1704 与病人皮肤 1701 之间的接触情况。图 17C 为一种机头的剖面图，其具有两个位于机头的一部分中的电触点以便使得
30 当元件 1704 与皮肤 1701 相接触时，接触件 1720 也与皮肤 1701 接触。接触情况可以通过测量接触件之间的电阻（或电容）来测定。当接触件 1720 之间的电阻（或电容）位于选定的范围（即皮肤常有的典型

范围)之内时,将会启动处理源 510。在另一个实施例中,接触件 1720 可为用于检测与皮肤表面 1701 的接触情况的磁传感器。在另外一个替代实施例中,接触件可为用于检测与皮肤表面 1701 的接触情况的机械传感器。例如,可以设置一个或更多弹簧针或按钮,以便使得当
5 元件 1704 与皮肤接触时,针或按钮就会压下。可以使用位于元件 1704 周围的多个传感器、针、按钮或其它机械传感器来帮助保证元件 1704 的全部表面与皮肤形成良好的接触。另外,接触件 1720 可为用于检测与皮肤表面 1701 的接触情况的常规型测压元件。可能优选使用容许测量电阻或电容的接触件、传感器、针、按钮或其它机械传感器来
10 保证与皮肤接触而非与其它表面接触,例如镜面或工作台面。

在另一个实施例中,使用一个或更多温度传感器来测定与皮肤表面 1701 的接触情况。典型皮肤温度范围为 30-32℃;相应地温度传感器可以位于与病人皮肤接触的装置的表面附近,并且当测量的温度处于选定范围(例如 23-27℃)内时可以确定发生接触。另外,当温度
15 传感器测量到指示接触温度与时间的关系曲线的斜率时,就可以确定已经发生接触。在另外一个实施例中,其中要将洗液分配于皮肤上(如上参照图 11 所述),可以利用位于喷口 1120 内的压力传感器来检测皮肤接触情况。压力传感器将会测量将洗液喷于皮肤上所需的压力。只有当机头与皮肤处于良好接触时,才会提供较高的压力来分配洗
20 液。

在标题为“用于光学射线机头的接触检测方法及设备”的由 Henry Zenzie 于 2001 年 4 月 30 日提交的美国专利申请 U. S. Application 09/847,043 中对接触传感器设计进行了更为详细的描述,其内容在此引入作为参考。

机头优选地在特定的速度范围内扫过病人皮肤。如果机头运动过慢(根据应用情况而定典型最小速度界限介于 5 至 25 毫米/秒之间),光照量将会过高因而可能会产生意外的热损坏。相应地,如果机头运动过快(根据应用情况而定典型最大速度界限介于 50 至 500 毫米/秒之间),光照量将会过低因而不能达到处理功效。因此,只有当机头
25 在此速度范围内扫过时,机头才会发射电磁射线进行处理。用于除毛/生长延缓的光美容机头工作的示例速度范围为 10-500 毫米/秒,该速度范围与大致等于剃刀经过其皮肤上的速度的速度范围相一致。
30

图 18A 为一种具有一用于测定机头速度的运动传感器 1820 的机头的一个实施例 1800 的剖面侧视图。通过向处理源（例如图 2 中的源 510）提供反馈控制，以便使得如果机头保持不动或者如果在皮肤 1810 上的运动过慢或过快，则可以分别降低或增加源的强度或者可关闭源，从而可以利用运动传感器 1820 来防止对皮肤 1810 造成伤害。任选地，处理源可以禁用，而不是减少功率。在一个实施例中，设置一轮子 1821 与皮肤 1810 形成物理接触，以便使得轮子随着机头 1800 相对于皮肤 1810 的运动而转动，因而可以测定机头速度。

机头 1800 可以适于当机头速度位于一可接受的速度范围之内或之外时通知操作者。例如，一触觉指示器（例如一振动器）可以适于当机头速度位于所要求的速度范围之内或之外时振动机头。另外，可以使用一视觉指示器（例如一发光二极管）或一听觉指示器（例如一报警器）来通知操作者机头速度位于所要求的范围之内或之外。在一些实施例中，可以使用多个指示器 1806（例如具有不同颜色的发光二极管，或者不同的声音指示器）来通知操作者机头速度过高或过低或者处于所要求的范围内。

图 18B 为一种具有至少一个轮子 1821 的运动传感器系统的一个实施例的一个实例的示意图。优选地，增加一第二轮子 1821 并且位于光学系统 206 的相对侧上以便保证光学系统 206 的全部皮肤接触表面以位于可接受范围内的速度运动从而在病人皮肤上提供均匀的照射。

在一个实施例中，每个外部轮子 1821 联接于对应的具有环绕着其周边的穿孔的辅助内部轮子 1822 上。一源 1830 沿着对应检测器 1832 的方向投射光以便使得当轮子 1821 转动时，辅助轮子 1822 的穿孔交替地传送和阻挡由源 1830 投射的光；因此，当机头 1800（在图 18A 中可看到）移过病人皮肤时，检测器 1832 就会产生一具有一串脉冲的信号。

普通技术人员将会理解，机头横过病人皮肤的速度与脉冲发生的重复频率成比例。一控制器 1834 使脉冲重复频率与机头速度相关联。上述带穿孔的辅助轮子的设计与标准计算机系统鼠标设计相似，例如由 Logitech of Fremont, CA 公司生产的 3 Bth Wheel Mouse 中的鼠标轮，其仅为用于测量机头速度的设备的一个实例，也可以使用许多其

它设备并且都在本发明的这个方面的范围之内。例如，在一个替代实施例中，一简单的电机联接于轮子 1821 上以便产生与机头速度成比例的电压。

图 19 示出了另一种具有一用于测定机头速度的运动传感器的光学设备 1900。在设备 1900 中，一光源 1902（例如一红外线发光二极管）联接于传送纤维 1904 中。一光检测器 1910（例如一便宜的 CCD 相机或二极管传感器）联接于接收纤维 1906 的端部。在设备 1900 中，传送纤维 1904 和接收纤维 1906 的端部联接在一起以便形成与皮肤 1908 相接触的单个纤维端 1909。一部分由传送纤维 1904 通过纤维端 1908 投射于皮肤表面 1908 上的光由皮肤表面 1908 反射或散射并由接收纤维 1906 通过纤维端 1909 接收并且由检测器 1910 进行检测。因为皮肤表面 1908 具有半周期性结构（例如，相似组织如毛囊、血管、腺之间的距离差不多为恒定结构），因此检测器输出以取决于机头速度的率进行调制。普通技术人员将会理解，机头速度可以由经过调制的检测器输出进行计算。任选地，可以增加通过纤维端 1911 联接在一起的第二传送纤维 1905 和接收纤维 1907，以便使得第一和第二传送纤维/接收纤维对位于光学系统 206 的相对两侧上，从而保证光学系统 206 的全部皮肤接触表面在可接受范围内移过皮肤以在病人皮肤上提供均匀的照射。

在系统 1900 中，每个传送纤维 1904、1905 联接于对应的接收纤维 1906、1907 上；另外，一传送纤维和对应的接收纤维可以在不同的分离点上与皮肤接触（即，传送纤维和对应的接收纤维不在皮肤处联接起来）；在这个实施例中，与皮肤接触的纤维的端部可以隔开任意的距离，该距离应使得由组织层散射的光子可以可靠地检测到。在这种实施例中，当联入接收纤维的光减少至散射的光子量所产生的信号太小以致不能准确测量时，就到达纤维间距的上限。

尽管已经对用于测量机头速度的光学设备进行了描述，但是应当理解其它速度测量方法也在本发明的这个方面的范围之内。例如，电磁设备通过在机头相对于皮肤运动时记录时间与皮肤的电（电容和电阻）/磁属性的关系曲线而测量机头速度。另外，可以测量在将一个物体拖过皮肤表面时所发射的声音的频谱和幅值，并且利用测得的信息来计算速度，因为声谱与速度相关。另一个替代方案是通过使用两个

沿机头顺着皮肤移动的方向隔开一段距离的传感器（例如一个在光学系统之前，而另一个在后），而利用热传感器来测量机头速度。在这种实施例中，第一传感器检测与机头速度无关的未处理过的皮肤的温度，而第二传感器检测照射后的皮肤温度；机头速度越低，输送至给定皮肤区域的能量密度就越高，就会导致第二检测器测量到越高的皮肤温度。因此，速度可以根据两个传感器的温度差进行计算。

一种利用热特征来测量机头速度的替代系统使用了一个位于沿机头顺着皮肤移动的方向与一热传感器隔开一段距离的位置处的热源（例如处理源或另一种加热皮肤区域的装置）。在这种实施例中，机头速度可以由热传感器所测量的温度来确定。对于低机头速度，热将有足够的时间来通过皮肤从热源传至热传感器；然而，在高速度下，热将没有时间到达热传感器。因此，由热传感器所测量的高皮肤温度将表示低速度，而低皮肤温度将表示高速度。

在速度传感器的一个替代实施例中，利用一光学设备来使用多普勒频移方法测量机头速度。在这个系统中，来自探针激光器的光波长被投射于皮肤上而速度通过光的反射部分的频移来确定。

在任一以上实施例中，速度传感器可与接触传感器（例如上文参照图 17A-17C 所述的接触传感器）协同使用。在机头的一个实施例中，接触情况和速度通过相同的组件来测定。例如，一种例如用于常规型计算机光学鼠标的光学鼠标型传感器可以既用来测定接触情况，又用来测定速度。在这样一个实施例中，使用了 CCD（或 CMOS）阵列传感器来连续地反映皮肤表面。如上所述，通过跟踪特定组的皮肤特征的速度，可以测量机头速度，并且由于当与皮肤接触时，由阵列传感器所接收的光学信号的强度就会增加，因此可以通过监测信号强度而测定接触情况。此外，可以使用一光学传感器例如 CCD 或 CMOS 装置来根据由皮肤反射回的光而检测和测量皮肤着色水平或皮肤类型；处理可以根据着色水平或皮肤类型而改变。

在本发明的一些实施例中，运动传感器与一反馈回路或检查表协同使用以便控制射线源输出。例如，所发射的激光功率可以根据检查表而与机头速度成比例增加。这样，可以在选定的深度保持固定的皮肤温度（即，通过在皮肤表面上保持恒定的通量），而不管机头是否在机头速度范围内运动。用来在特定深度达到给定皮肤温度的功率在

美国专利申请 U. S. Pat. Application No. 09/634,981 中进行了更详细的描述，其在上文中已引入本文作为参考。另外，也可以检测处理后皮肤温度，并且利用一反馈回路来通过改变激光输出功率而在皮肤表面上保持基本恒定的能量密度。皮肤温度可以使用其它常规型热传感器或非接触式中红外线光学传感器来检测。以上运动传感器为示例性；运动检测可以通过其它方法例如声音来实现（例如使用多普勒信息）。

尽管以上实施例参照一种在由操作者移动时监测机头速度的系统进行了讨论，但机头可以安装于一移动台上以便使机头按照受控的预定速度移过皮肤表面。在这样一个实施例中，设备将会相对于病人进行定位以便处理选定的皮肤区域，并且移动台可以根据需要移至下一区域。

图 20 为一种机头的一个实施例 2000 的一个实例的示意图，示出了一种独立式光美容装置的一些方面。机头 2000 包括一光源 2055、一电源 2047、一光学系统 2044、一冷却系统 2046 以及一速度和/或接触传感器 2048。所示的装置与皮肤区域 2043 相接触。光学系统 2044 将光从光源 2055 联接至皮肤处理区域 2043 中。

冷却系统 2046 可为一相变冷却器或任意其它适用的冷却系统。在一些实施例中，冷却系统 2046 与散热片 2045（或者电极或其它冷却表面，未示出）形成良好的热接触。电源 2047（例如电池或电容器）向光源 2055 供应电流。接触和/或速度传感器 2048 保证了处理安全有效，如上文所述。尽管所示的接触和速度传感器为单一组件，但是应当理解接触和速度传感器可为不同的组件，并且可有多于一个每种类型的传感器，如上所述。控制电子电路 2049 处理来自接触/速度传感器 2048 或其它传感器（例如热传感器）的数据并且控制着光源 2055 和冷却系统 2046。冷却系统 2046 可在处理之前通过热接触板 2050 进行冷却。电源 2047 可以通过电触点 2051 充电。开关按钮 2052 控制着电源。壳体 2053 可用来封装、保护或者安装一个或更多上述部分。

任选地，可设置一除毛装置 2054，用来在利用来自光源 2055 的光照射之前进行除毛，以便保证基本上没有毛发在皮肤表面上延伸。例如，除毛装置 2054 可为一剃刀片（例如一安全剃刀、一夹头剃刀）、一电动剃刀、一将毛发粘附于一表面上并且在机头移过用户皮肤上时拔出的脱除装置（例如类似于由 Happy Lady Inc. 公司生产的

Epilady™ 的装置)、一用于研磨毛发的研磨装置、或者一用于溶解毛发的化学化合物。除毛装置可为一次性以便使得除毛装置易于由用户进行更换。在粗毛的情况中,可以使用具有一个或多个刀片的剃刀;然而,在细毛的情况中,可以使用一研磨纸。具有粗毛的身体位置首先可以经过一次或更多次光美容处理之后变为具有细毛;因此,头几次处理可以使用剃刀片而随后的处理可以使用研磨纸。在一些实施例中,研磨纸可以随着光美容装置的一个行程而简单地移过皮肤,而在其它实施例中,研磨纸可以通过一振动机构(例如一马达)而振动。

图 21 为一种用于停放机头 2000 的机头停放台的一个实施例 2100 的一个实例的示意图。停放台 2100 容放于壳体 2155 中。电源 2156 通过电触点 2051 而对电池/电容器 2047 充电。冷却材料 2046 由冷却器 2157 (例如一 Peltier 元件) 进行冷却。例如,冷却器 2157 可通过凝结一相变液体或冻结一相变固体而补给冷却系统。散热片 2058 散发由冷却器 2157 所产生的热。散热片 2058 可以使用气体、液体或固体(相变)介质来除热或者仅为通过暴露于室温下来进行冷却的翼片。连接电缆 2159 容放着用于从一电出口向停放台供应电功率的导线,并且还可以包括用于对散热片 2058 进行水冷却的管道。在 G. Altshuler 等人于 2000 年 12 月 28 日提交的标题为“用于 EMR 处理的方法及设备”的美国申请 60/292827 中对一种独立式光美容装置和一种机头停放台进行了更详细的描述,其内容在此引入作为参考。

对于光美容装置的一些实施例,具有一个或更多可更换组件将很有利。例如,在一些实施例中,其中机头可能会被掉落或误用,这时使得一个或更多光学系统可从机头上拆下可能会很有利。此外,为了进行各自需要不同的光源或光学系统的各种处理(例如着色病变去除处理和除毛处理),可互换的光学元件将使得用户能够利用相同的机头执行不同的应用。此外,对于采用寿命有限的光源或电源的系统来说,可能需要在使用寿命结束时更换光源。

图 22 为一种具有一可分离式头部 2210 的机头的一个实施例 2200 的一个实例的示意图。机头 2200 具有一联接于头部 2210 上的手柄 2220。手柄 2220 可利用任意已知的紧固方法联接于头部 2210 上。优选地,头部 2210 包括便于使用可更换组件的光学组件(例如图 16A 的头部 1600)。

图 23 为一种模块式机头的一个实施例 2300 的一个实例的示意图，其具有一个或更多适于易于制造并且/或者易于用户更换的组件。例如，机头 2300 便于头部部件 2310(包括一光学系统)、冷却部件 2320 以及电源部件 2330 的装配和/或更换。优选地，模块式机头 2300 适于使得当已装配好时，头部部件 2310 与电源部件 2330 的相配电源插头相接触。

图 24 为一示意图，示出了一种包括一源 2410(例如两个二极管激光棒)的光学部件 2400。源 2410 可以并入一用户可更换的一次性筒中，其包括电极 2412、散热片 2430、光学系统 2420 和联接板 2440。联接板 2440 可用于紧固光学系统 2420、源 2410 和散热片 2430。优选地，源 2410 的紧固机构适于自动将源 2410 与光学系统 2420 对齐。同样优选地，联接板由具有良好导热性的材料(例如铜)制成以便从光学系统 2420 导热。为了简化源 2410 与元件 2420 的对齐，源 2412 可以固定安装于光学系统 2420 上。

除了在其使用寿命结束时更换源 2410 之外，还可能需要便于用户更换源 2410 以便用于不同的美容处理而不须购买多个机头。另外，还可能需要便于用户根据皮肤类型、毛发类型和/或待处理的皮肤区域的位置(例如腋下、三角区、腿、面部)而更换光源 2410。

图 25 为一种光美容装置 2500 的一个实例的示意图，示出了本发明的一些方面。装置 2500 具有一头部 2580 和一手柄 2590。头部 2580 具有用于形成第一射线区域(例如图 3 中的区域 311)的第一光学系统 2510(例如图 3 中的光学系统 310)，以及用于在病人皮肤上形成第二射线区域(例如图 3 中的区域 316)的第二光学系统 2515(例如图 3 中的光学系统 315)。如上文参照图 3 所述，用于形成第一区域和第二区域的射线可以来自单个分开的源或两个源(未示出的源)。装置 2500 还包括一运动传感器系统，该运动传感器系统具有一轮子 2521(例如对应于图 18 的轮子 1821)和一第二轮子 2522(例如对应于图 18 的轮子 1822)，第二轮子 1822 位于光学系统 2510 的相对侧上以便保证光学元件 2510 的全部皮肤接触表面以位于可接受范围内的速度运动从而在病人皮肤上提供基本均匀的照射。

图 26A 为一种光美容头部 2600 的一个实例的示意图，示出了针对处理皮肤的弯曲区域(例如颈、背或臂)的本发明的方面。头部 2600

包括两个用于输送电磁射线的枢轴旋转输送系统 2610 和 2620。头部 2600 的组件基本上容放于一壳体 2630 内并且通过软线 2640 联接于基本单元（未示出）上。为便于描述，壳体 2630 示出为透明线框的形式。头部 2600 的组件的尺寸可以根据它们将要用于的身体部分而选定，并且多个头部可以连接于软线 2640 以便容许处理不同的身体部分。另外，每个头部也可以具有一固定的软线以便使得每根软线能够插于基本单元中并且能够移除。

图 26B 为用于处理弯曲表面的一种头部的双传送系统 2610 和 2620 的一个实施例的示意图。为了示出它们的相对位置关系，所示的传送系统 2610 和 2620 不带壳体。图 26B 示出的传送系统沿着至少一个旋转方向作枢轴旋转以便于与皮肤的弯曲区域保持接触。例如，传送系统 2610 和 2620 可以相对于彼此成一定角度（例如 5-30 度）安装并且安装方式使得能够绕着轴线 X 和 X' 旋转。

图 27 为一示意图，示出了根据本发明的机头 2700 的一些方面的一个实施例。机头 2700 包括一具有手柄 2702 和头部 2704 的壳体 2710。机头 2700 包括一头部部件 2710（包括一光学系统）、一冷却部件 2720 以及一电源部件 2730。

图 28 为根据本发明的至少一些方面的光美容装置的一个实施例 2800 的示意图。装置 2800 包括一机头 2810、一基本单元 2820、一将机头 2810 联接于基本单元 2820 上的软线 2826。机头 2810 可以由操作者抓住以便将头部 2830 移过病人皮肤（未示出）。头部 2830 可为任一种上文中所述的头部或者任一种其它适用的头部以便实现光美容处理，例如下文中所述的任一种处理。

以下讨论了可以利用根据本发明的设备及方法实现的处理的实例；然而，所讨论的处理仅为示例性而非进行限制。根据本发明的设备及方法有很多种并且可以用于任意已知或尚待开发的处理。

示例性的处理机理包括由位于导致不必要的美容条件的组织内的发色团或者由接近该组织的发色团来吸收光。处理可以通过在造成不可复原损害的温度之下对目标组织进行有限加热来实现，或者可以通过加热以产生不可复原的损害（例如使变性）来实现。处理可以通过直接刺激对热的生物响应而实现，或者通过诱导一连串现象以便由热间接得到生物响应而实现。处理通过任意上述机理的组合而产生。任

选地，可以对处理区域或邻近区域应用冷却、DC 或 AC (RF) 电流、物理振动或其它物理刺激/作用以便增加处理功效。处理可以通过单个处理期来实现，或者可以使用多个处理期以便达到所需的临床效果。

5 根据本发明的一个或更多方面的装置可以在各种光学范围内操作。例如，输送至皮肤的电磁射线可以具有 380 - 1900 纳米范围内的波长。所输送的光的功率可以在 0.001-300 瓦/厘米的范围内，而示例性的扫描速度包括 0.1-500 毫米/秒。所需的射线特征可以通过适用的发光二极管、灯和二极管激光棒或其它现有的或尚待开发的适用光源来实现。

10 射线诱导的除毛为一种可通过根据本发明的方面的设备及方法执行的美容处理。在除毛的情况下，热破坏的主要目标是毛球，并且优选球的毛基质、毛乳头或基底膜。对于除毛处理，位于毛干和囊中的黑色素为目标发色团。尽管球包含黑色素因而可进行热处理，但在球内的乳头与毛干内的基质之间提供了毛发生长通道的基底膜中包含着
15 最高浓度的黑色素并且可选择作为目标。

介于 0.6 至 1.2 微米之间的波长通常用于除毛。通过对功率、速度和聚焦几何进行适当组合，可以将不同的毛发相关的目标（例如球、基质、基底膜、干细胞）加热至变性温度而周围的真皮保持不会受到损害。由于目标毛囊和表皮都包含黑色素，因此可以利用表皮接触冷
20 却和长脉冲宽度的组合来防止发生表皮损害。在由 Rox Anderson 等人于 2002 年 3 月 12 日提交的标题为“用于控制毛发生长的方法及设备”的共同未决的临时专利申请号 60/363,871 中给出了对除毛的更详细说明，其在此引入作为参考。

除毛通常需要覆盖大面积（例如背和腿），因而为实现短处理时间，所需的功率相应地也很大（大约 20 - 500 瓦）。当代的二极管棒能够在 800 纳米下发射 40 - 60 瓦的功率，这就使得它们能够有效用于根据本发明的光美容装置的一些实施例中。

毛发生长处理的示例性方法可以通过组合利用光对毛囊进行低功率照射并物理拔出毛干，以及/或者从体内完全或者不完全地物理拔出
30 毛囊来实现。根据一些实施例，照射作用通过利用发射可由囊内的黑色素或者其它内生或外生发色团吸收的波长范围的光的光源来照射包含毛囊的皮肤的一部分来实现。物理拔出作用可以通过机械、机电或

其它适用方法而进行。这种处理可以用于临时性减毛或永久性减毛。

根据本发明的一种毛发生长处理方法的一个第一示例性实施例包括首先物理上除毛（“脱毛”）然后按照上述照射皮肤。根据一些实施例，除毛可以调整为主要从毛囊中除去毛干；另外，除毛也可以下至
5 角质带。这种脱毛可以通过机电脱毛或者上蜡来进行。

可以利用例如上述光美容装置的实施例之一来进行光处理。根据这些实施例，光由毛基质中的黑色素吸收并且作为热损伤的结果，毛发生长受到减缓或者受到完全抑制。

任选地，在脱毛之后但在照射之前，可以将洗液局部应用于处理
10 区域中的皮肤上（例如通过机头）以便充满由于除毛而变空的毛囊。在一些实施例中，所选定的透明洗液具有的折射率所处的范围适于提供波导效应以便将光引至待照射的皮肤区域。优选地，洗液的折射率高于水的折射率（即，根据水的化学添加剂而定大约为 1.33）。在一些实施例中，洗液的折射率高于真皮的折射率（即，大约为 1.4）。在
15 一些实施例中，洗液的折射率高于内根鞘的折射率（即，大约为 1.55）。在折射率高于内根鞘的折射率的实施例中，入射在皮肤表面上的光可以直接输送至毛基质而不会衰减。

用来照射皮肤的有效脉冲长度由束尺寸除以照射源的扫描速度给定。例如，以 50 - 100 毫米/秒的扫描速度移动的 2 毫米的束尺寸提供
20 的有效脉冲长度为 20 - 60 毫秒。对于 250 瓦/厘米的功率密度，有效能量密度为 5 - 10 焦耳/平方厘米，大约两倍于在不使用高折射率洗液的情况下装置所输送的光的能量密度。

在一些实施例中，可以调节洗液的 pH 值以便降低基质细胞的变性阈值。在这些实施例中，只需要较低的功率来损伤毛基质从而提供
25 毛发生长处理。任选地，洗液可以掺加以能够显著吸收由源所发射的光的分子或离子或原子。由于通过洗液的作用而使得毛囊中光吸收量增加，因此就可以使用较低功率的照射源来提供足够的射线用以加热毛基质。

根据本发明的一种毛发生长处理方法的第二示例性实施例包括首先
30 照射皮肤，然后物理上除毛，如上所述。通过首先照射皮肤，可以削弱毛干与囊的连接或者毛囊与真皮的连接。因此，机械或机电法脱毛就可以更容易实现（例如，通过使用软上蜡或机电脱毛器）并且可

以减轻疼痛。

照射可以削弱毛球与皮肤或皮下脂肪的连接；因此，与单独使用脱毛相比，就可以从皮肤中拔出高得多比例的毛囊。由于毛球的直径接近外根鞘的直径，因此拔出毛及毛球就可以永久性地破坏整个毛囊包括干细胞。相应地，通过首先照射然后脱毛，就可以延缓或者终止新的毛发生长。

脂肪团处理为可以通过根据本发明的方面的设备及方法来处理的美容问题的另一个实例。特征脂肪团凹窝的形成从不良血液和淋巴循环开始，而这又抑制了细胞废物的排除。例如，细胞内空间的未排除死细胞可能会随着时间的推移而渗漏脂类。由于毒素和细胞废物的不断积累，就会造成结缔组织损坏和随后的结核形成。

以下为两种示例性的脂肪团处理，两者的目的都在于既刺激血流和又刺激纤维原细胞生长。在第一示例性处理中，利用发射近红外光谱范围（例如 650 - 1850 纳米的波长范围）的处理源与设计用于聚焦于皮肤表面下方 2 - 10 毫米的光学系统的组合来产生局部的热损坏区域。在一个实施例中，将功率密度为 1 - 100 瓦/厘米的光输送至皮肤表面，并且按照一定速度操作设备以便在皮肤下方 5 毫米的距离处产生 45 摄氏度的温度。可以施加冷却作用以避免或减少对表皮的损坏从而减少形成伤口。在皮肤下方的选定距离处产生选定温度的更多细节在 2000 年 8 月 9 日提交的美国专利申请 U. S. Patent Application 09/634,691 中给出，其内容已在上文中引入作为参考。处理可以包括压缩组织、按摩组织、或者多次通过组织之上。

在第二示例性处理中，利用发射近红外光（例如发射波长范围为 700 - 1300 纳米的发光二极管）的处理源来将光聚焦于皮肤表面下方 2 - 10 毫米的距离处，并且将真皮/皮下脂肪温度提高至适当位于热损坏阈值之下的点（例如 42 - 60 摄氏度范围内的温度）。根据第二示例性处理，加热作用可以提高脂解（即脂肪破坏）率并引起脂肪细胞的凋亡（即，拟定的细胞死亡）。任选地，局部的脂解膏可以与第二示例性处理组合使用；在真皮/皮下脂肪中的升高的温度分布图可以增强膏的穿透力从而提高其功效。由于皮下脂肪的热松弛时间很长（即长于 1 分钟），对一个区域进行多次扫描处理就可以实现所需的对脂肪的加热作用，而同时保持正常的皮肤表面温度。以上示例性处理可以

用于脂肪新陈代谢激活和减脂处理。

粉刺为可以利用根据本发明的方面的设备及方法来处理的另一种非常常见的皮肤病。当来自皮脂腺的皮脂不能通过毛囊到达皮肤表面时就会产生粉刺，并且会在毛囊内造成细菌感染。光美容处理为传统
5 处理方法（例如敷服或口服）的替代方案。

以下为根据本发明处理粉刺的示例性方法。在每一种示例性方法中，实际的处理区域可以比较小（假定处理的是面部粉刺），因此可以使用低功率的 CW 源。第一可能的处理为有选择地损坏皮脂腺以防止皮脂产生。皮脂腺位于皮肤表面以下大约 1 毫米处。通过在这个深度处产生一焦斑并且使用由脂类有选择地吸收的波长（例如接近于
10 0.92、1.2 和 1.7 微米），直接热破坏就变得可能。例如，为了造成热变性，可以使用 2000 年 8 月 9 日提交的美国专利申请 U. S. Patent Application 09/634,691 中所述的任一种方法在皮肤表面下方大约 1 毫米处产生 45-65 摄氏度的温度，该申请的内容已在上文中引入作为
15 参考。

任选地，可以使用焦斑线性矩阵（如上文参照图 4 所述）来产生多个损坏岛。尽管皮脂腺的准确位置可能并不知，但是每次利用焦斑矩阵来进行处理都将会造成一定数量的皮脂腺受到损坏。这样，通过多次对区域进行处理，大量的皮脂腺将会受到损坏。

粉刺的一种替代处理包括将皮脂腺加热至位于热变性温度以下的某个点（例如，加热至 45-65 摄氏度）以便实现终止皮脂产生和凋亡（即，拟定的细胞死亡）。这种有选择的处理可利用负责皮脂产生的细胞相对于周围细胞的低热阈值。粉刺的另一种替代处理为对通向皮脂腺的血液供应进行热破坏（例如，通过加热血液至 60-95 摄氏度的温度）。
20
25

对于以上对粉刺的各种处理，可通过使用化合物例如靛青绿（ICG，吸收 800 纳米附近）或亚甲蓝（吸收 630 纳米附近）来使得皮脂腺对近红外射线敏感。另外，也可使用非热光动力治疗剂例如 photofrin 来敏化皮脂腺。在一些实施例中，可以使用生物化学载体例如单克隆抗体（MABs）来有选择地将这些敏化化合物直接输送至皮脂腺。
30

尽管以上程序被描述为用于处理粉刺，但由于这些处理涉及损坏/

破坏皮脂腺（从而减少皮脂输出），因而这些处理也可用来处理过多油性的皮肤。

另一种处理粉刺的光基方法包括对造成与粉刺相关的特征炎症的细菌（P. 粉刺）进行热破坏。对细菌的破坏可通过瞄准存储于 P. 粉刺中的卟啉而实现。卟啉，例如原卟啉、粪卟啉、以及锌-原卟啉由厌氧细菌作为其新陈代谢产品而合成。卟啉吸收 400 - 700 纳米的可见光谱区域内的光，最大吸收能力在 415 纳米左右。通过提供足够强度的选定波长范围的光，通过吸收而产生的热就会造成细菌死亡。例如，使用设计用于聚焦于皮肤表面下方 0.2 - 1 毫米的光学系统和在皮肤表面上为 0.01 - 10W/cm 的功率密度，利用发射波长范围为 360 - 700 纳米的处理源，就可以达到所需的效果。

另外一种用于处理粉刺的技术包括使用光来扩张受到感染的毛囊的开口以便容许皮脂无阻碍地流出。在这种技术的一个实施例，将优先聚集于囊开口中的洗液（例如带有有机无机染料或吸收粒子的脂类稠洗液）施加于皮肤表面上。处理源波长与洗液的吸收带相匹配。例如，在掺加 ICG 的洗液的情况中，源波长为 790 - 810 纳米。通过使用光学系统来在漏斗管/下漏斗管处产生 45 - 100 摄氏度的温度，例如，通过在皮肤表面上产生一定能量密度（例如 1 - 100 瓦/厘米），囊开口就可以扩张并且容许皮脂流出毛囊并重造下漏斗管以便防止黑头粉刺（黑头）形成。

非消融性皱纹处理现在已用作传统的消融性 CO₂ 激光皮肤重修面处理的一种替代方案，其为可通过根据本发明的方面的设备及方法来执行的另一种美容处理。非消融性皱纹处理通过同时地冷却表皮并将光输送至真皮的上层以便对纤维原细胞进行热刺激从而产生新的胶原沉积而实现。

在皱纹处理过程中，由于主发色团为水，因此 0.8 - 2 微米的波长范围为处理射线的适用波长。由于通常只有面部的皱纹具有美容意义，因此处理的区域通常比较小并且所需的覆盖率（平方厘米/秒）因而也很低，并且可以使用较低功率的处理源。可以组合使用提供表面下聚焦的光学系统与表皮冷却作用来达到所需的结果。真皮上方温度的精确控制很重要；如果温度过高，对表皮产生的热损坏将会过大，而如果温度过低，新的胶原沉积量将会很少。可以使用一速度传感器

(在手动扫描机头的情况中)或者一机械驱动器来精确控制真皮上方温度。另外,也可以使用一非接触式中-红外热传感器来监控真皮温度。

5 血管病变(例如葡萄酒色痣、红斑痤疮、蜘蛛脉)为可利用根据本发明的方面的设备及方法来处理的另一种美容问题。对于血管病变的处理,目标发色团为这些病变中的血液。对于表面血管疾病,示例性的处理波长范围为0.4-0.6微米,而对于深层血管病变,则为0.6-1.3微米。在处理蜘蛛脉的情况中,目标组织的较大尺寸和相应较长的热松弛时间要求长时间地大量沉积能量以便实现热破坏并保护表皮。此外,可利用主动表皮冷却作用(特别是对于具有较的皮肤类型IV-VI的病人)来防止造成表皮损坏。在病变的处理中,使用CW源特别有利,因为与除毛类似,部分目标结构(脉壁)中包含很少的血液并且必须通过热扩散来损坏。

15 着色病变例如年龄斑可以通过有选择地以这些结构中的包含黑色素的细胞为目标而除去。这些病变利用聚焦于皮肤表面之下100-200微米深度的光学系统来定位并且可以利用0.4-1.1微米范围的波长来瞄准。由于各个单独的带有黑色素的细胞很小并且热松弛时间很短,因而较浅的表面下聚焦有助于达到变性温度。

20 腋臭的消除为可利用根据本发明的方面的设备及方法来处理的另一种问题。在这种处理中,使用一种具有由小汗腺/顶泌腺有选择地吸收的波长的源来对小汗腺/顶泌腺进行热损坏。任选地,可使用敏化化合物来加强损坏。

25 纹身去除为可利用根据本发明的方面的设备及方法来处理的另一种程序。用于纹身去除的常规型装置包括用于美容纹身处理的短脉冲(10-50纳秒)Q交换的红宝石、紫翠玉、钕:钇铝石榴石和倍频钕:钇铝石榴石。通常,源波长根据待去除的纹身的颜色来选择(例如,利用绿色激光来除去纹身的红色部分)。由于油墨粒子实际上已合入各个细胞中,因而用于纹身去除的热处理的一个实施例会造成细胞破裂,从而释放油墨。

30 用于纹身去除的根据本发明的方面的设备的示例性实施例利用一CW源和一将来自处理源的射线紧密聚焦于包含油墨粒子的细胞所居留的深度(150-700微米)的光学系统来使包含油墨的细胞破裂。另

外,也可以将这些细胞加热至其热变性点之下并且产生凋亡。在设计用于产生凋亡的实施例的情况中,通过在准连续模式下操作射线源同时将机头连续地扫过皮肤表面以便形成细胞受到损坏的区域和在中间具有未经照射区域的区域,就可以加强复原效果。在一些实施例中,可以使用来自速度传感器的反馈来控制激光发射并形成与机头速度无关的等间距的损坏线。为了完全除去纹身,可能需要多次处理。

在一些常规型的比较昂贵的纹身去除设备中,一发射 0.532 微米波长的 Q 交换的倍频钕: 钇铝石榴石激光器与一发射 1.064 微米波长的 (钕: 钇铝石榴石)、一发射 0.755 微米波长的紫翠玉激光器组合使用; 各个激光器有选择地操作以便瞄准包含不同纹身油墨颜色的细胞。根据本发明的方面的模块式设备的实施例提供了以上系统的较低成本替代方案。例如,本发明的一个实施例可以适于容许使用多个发射不同波长或波长段的光源或者单个光源和用于改变由源产生的光的波长的光学组件。特别地,为了达到接近 0.755 微米的波长,可以使用一 0.808 微米的二极管激光棒; 并且可将一钕: 钇铝石榴石晶体模块插入由二极管激光棒装满的机头中以便产生接近于 1.064 微米的波长; 并且为了产生接近于 0.532 微米的波长,可以使用一个二次谐波发生晶体来将发射 1.064 微米波长射线的激光二极管的频率加倍。另外,可以使用一自倍频晶体例如 Nd: YCOB。

低强度治疗 (LIT) 为可利用根据本发明的方面的设备及方法来实现的另一种程序。LIT 可用于处理伤口、腕管综合症处理、或者用于刺激毛发生长、或者用于加速生物化学反应。通常用于 LIT 的功率密度和波长 (630 - 820 纳米) 可以利用二极管激光器或发光二极管处理源来实现。任选地,一次或更多次以上处理可以用于兽医 LIT 应用中。

突出的伸长标志和疤痕的消除或减少为可利用根据本发明的方面的设备及方法来实现的程序。与非消融性皮肤重修面的情况类似,为了实现以上程序,可以通过在上真皮中形成一个薄的热损坏层来刺激胶原沉积和伤口复原效果。

疣的去除为可利用根据本发明的方面的设备及方法来实现的另一种程序。疣的去除可以使用一产生的光位于血液吸收区域 (0.5 - 0.8 微米) 中的源来实现。这个波长由血红蛋白有选择地吸收,其好像切

断了对疣的血液供应。

牛皮癣为可利用根据本发明的方面的设备及方法来处理的另一种皮肤病。适于处理牛皮癣的本发明的示例性实施例发射接近 800 纳米的波长。任选地,可以使用一种或更多种敏化剂例如光动力药物或 ICG/ 5 亚甲蓝。每周可以应用若干次处理,并且可以按照若干种不同方式输送,包括处理岛(或线)。根据本发明的方面的设备及方法的其它应用包括助于将敷药和美容制剂输送至皮肤中。

这样就对本发明的概念和许多示例性实施例进行了描述,但是本发明所述领域的普通技术人员应当清楚,本发明可以按照各种不同的方式来实施,并且这些技术人员很容易进行变动和改进。因此,所给 10 出的实例并非用于进行限制。本发明只是根据需要由以下权利要求及其等价内容来限定。本发明只是根据需要由以下权利要求及其等价内容来限定。另外,应当理解,使用的词“包括”或者“具有”意味着包括其后列出的项和其等价内容以及所列的这些项之前、之后或者之 15 间的其它项。

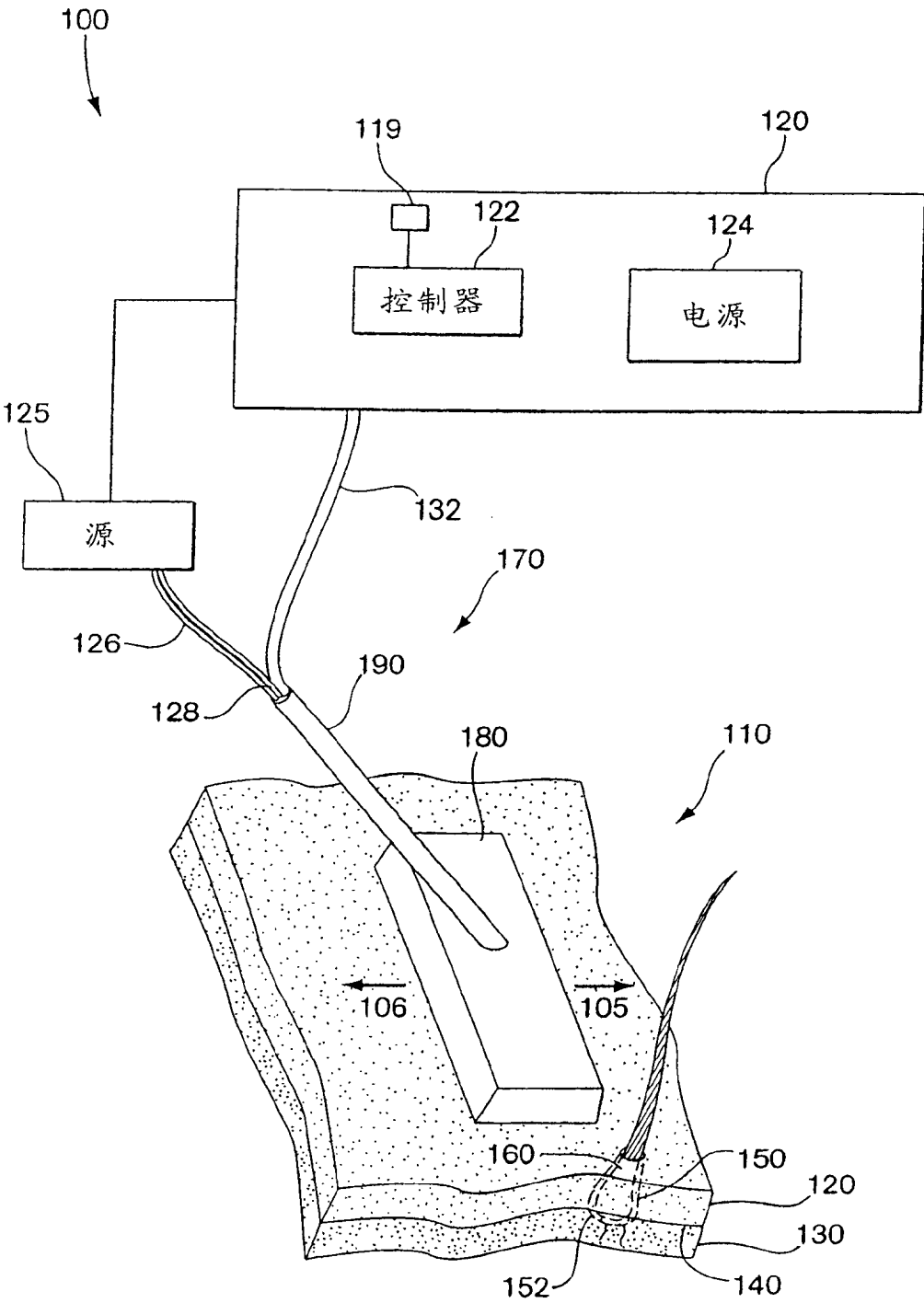


图 1

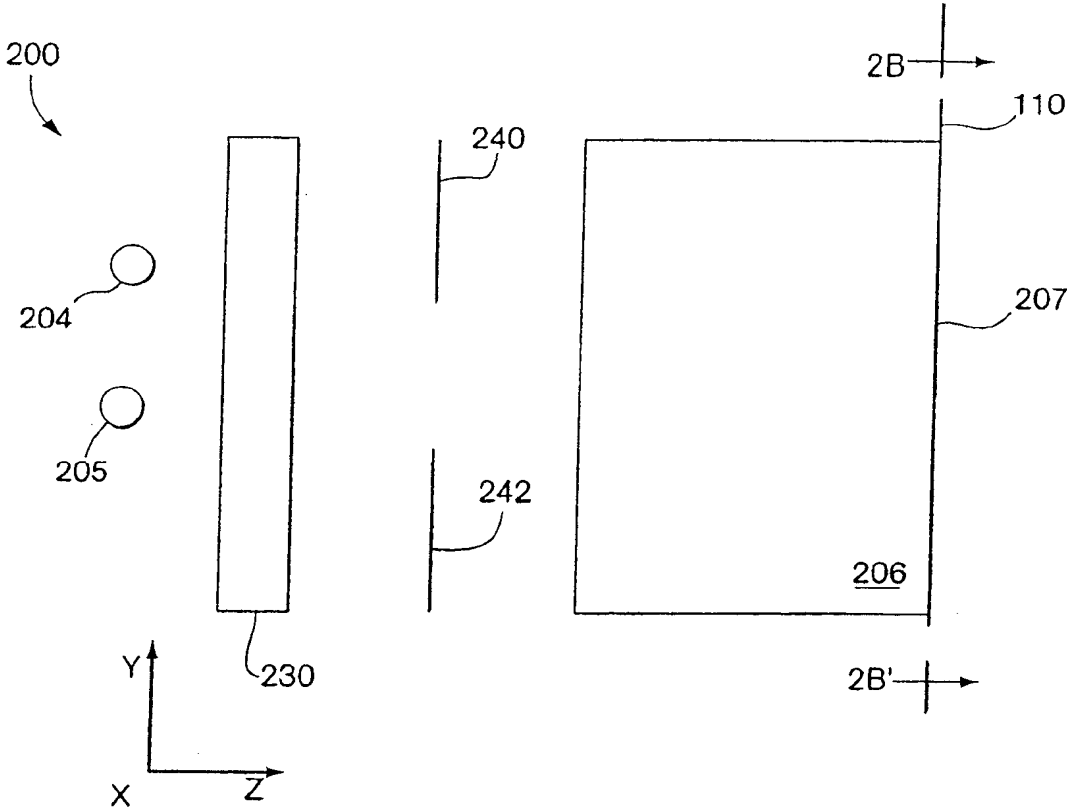


图 2A

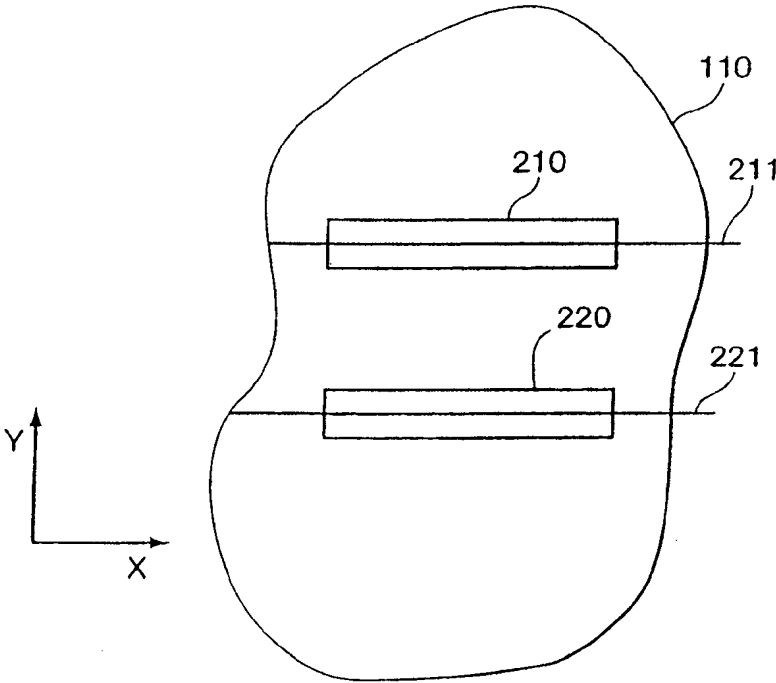


图 2B

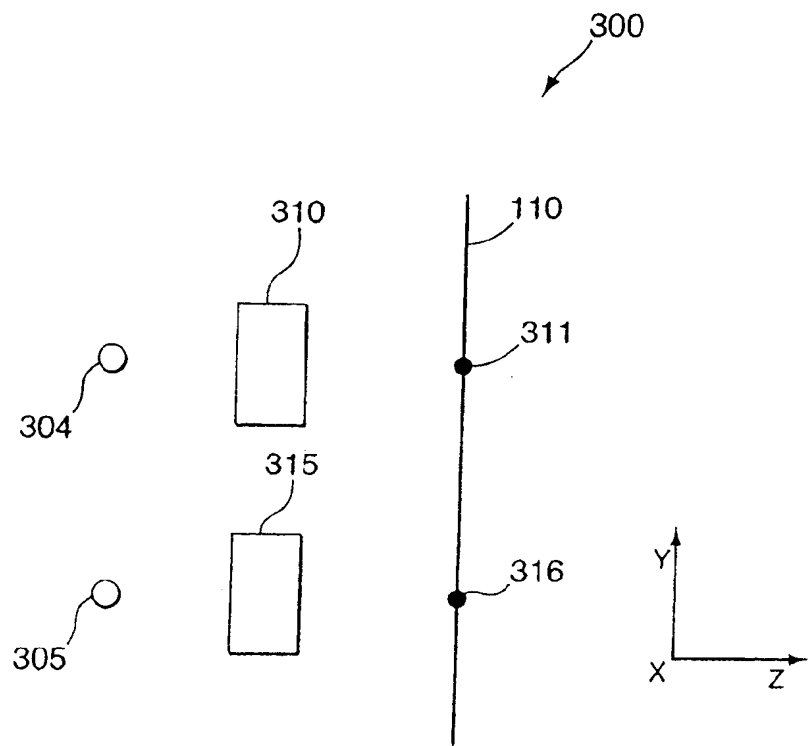


图 3

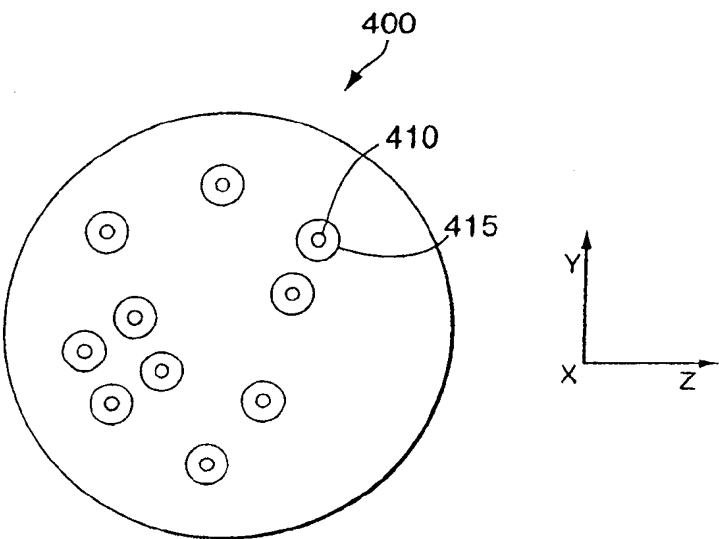


图 4

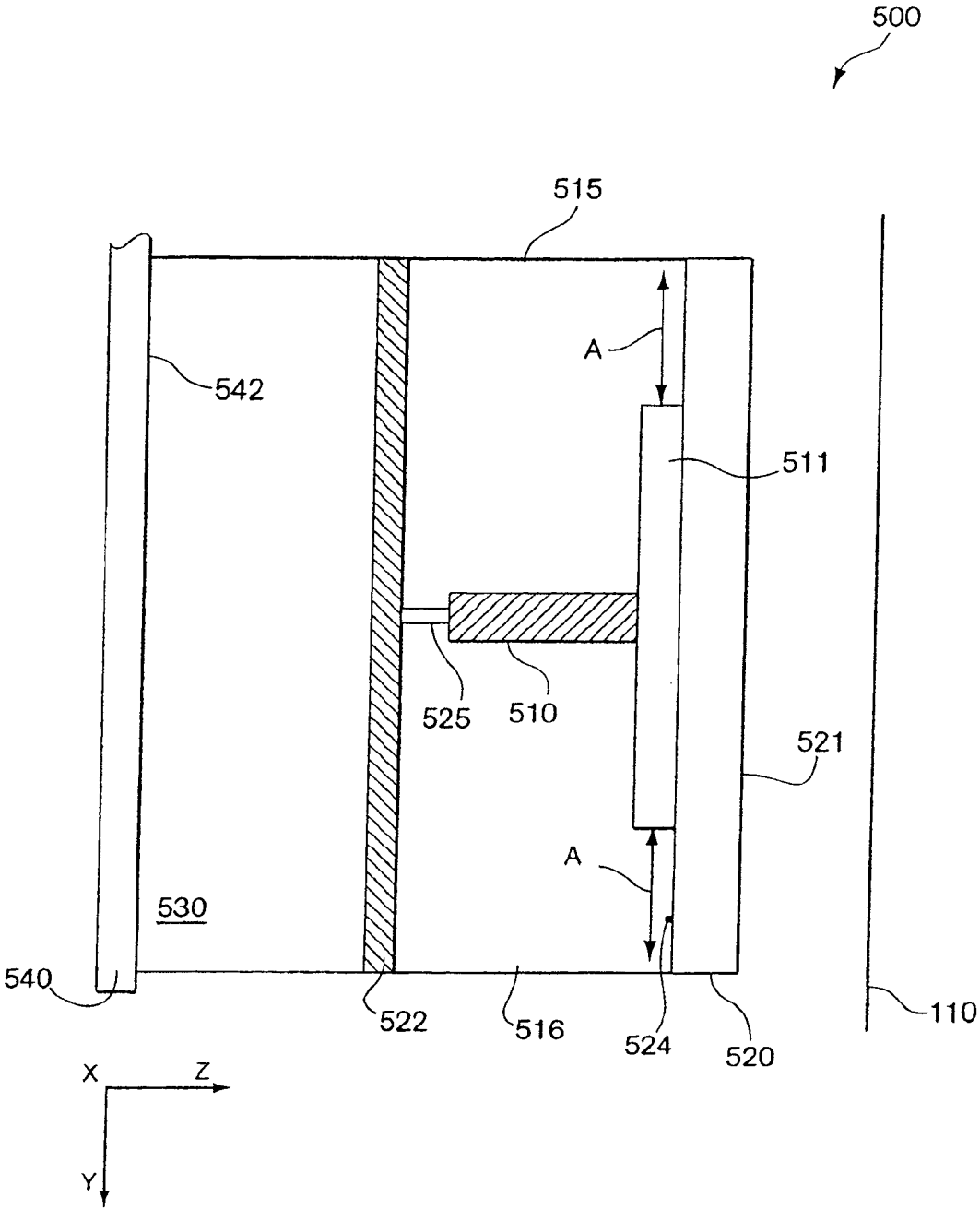


图 5

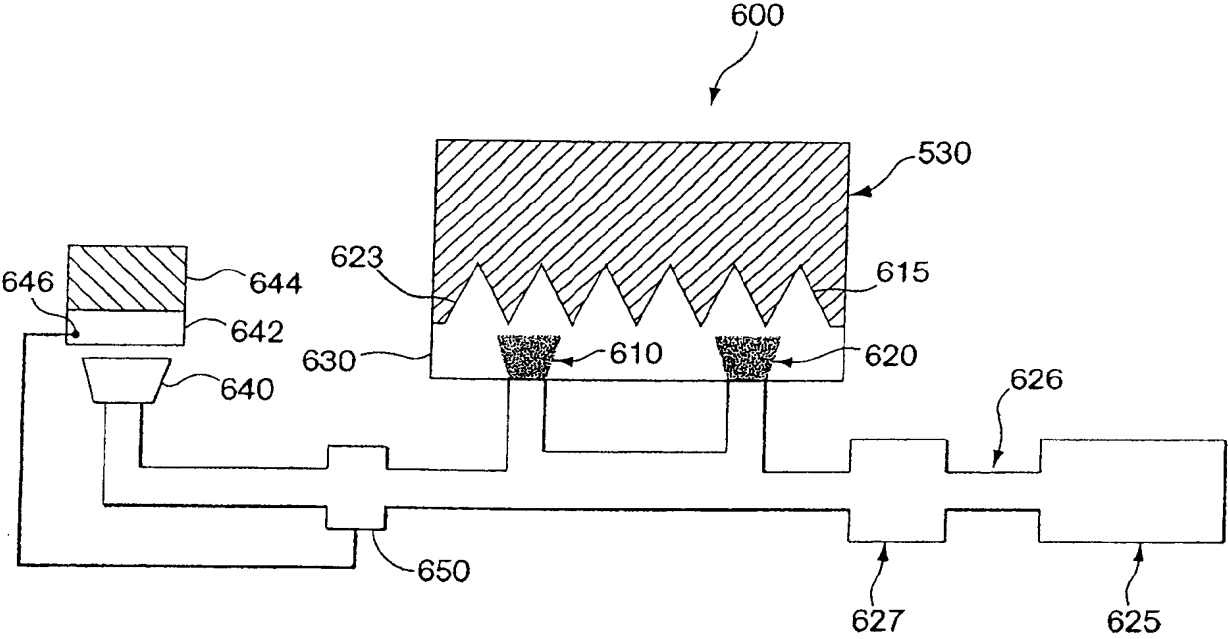


图 6A

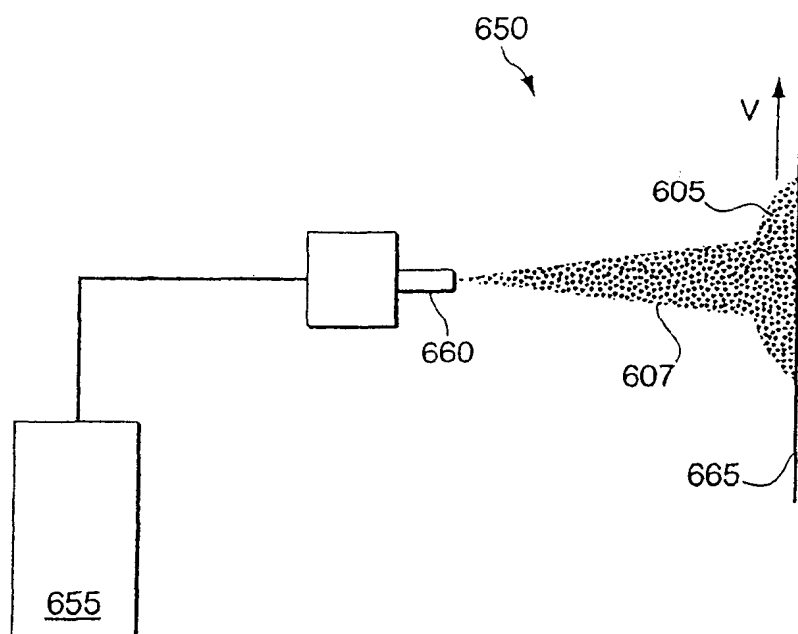


图 6B

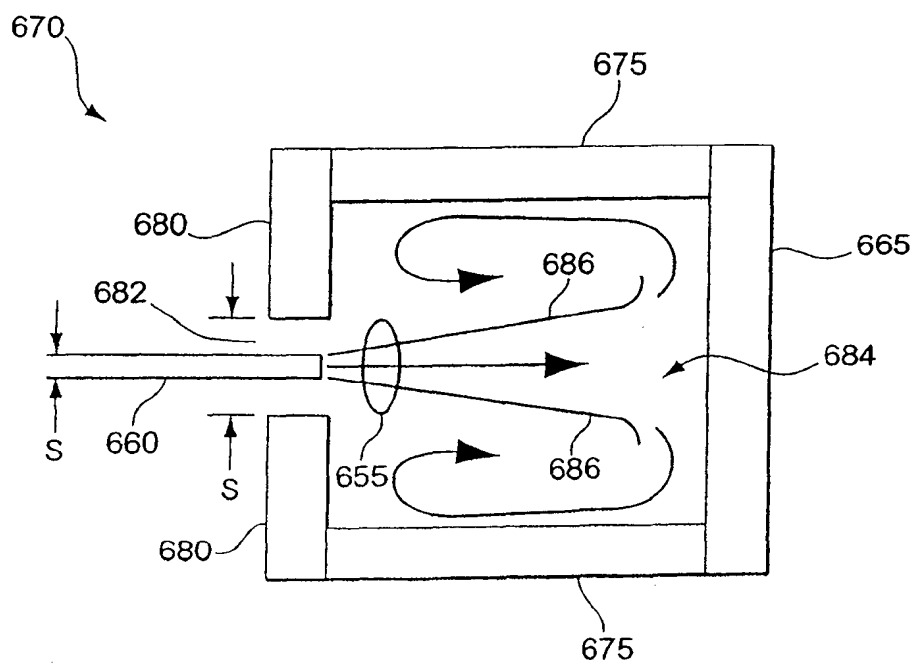


图 6C

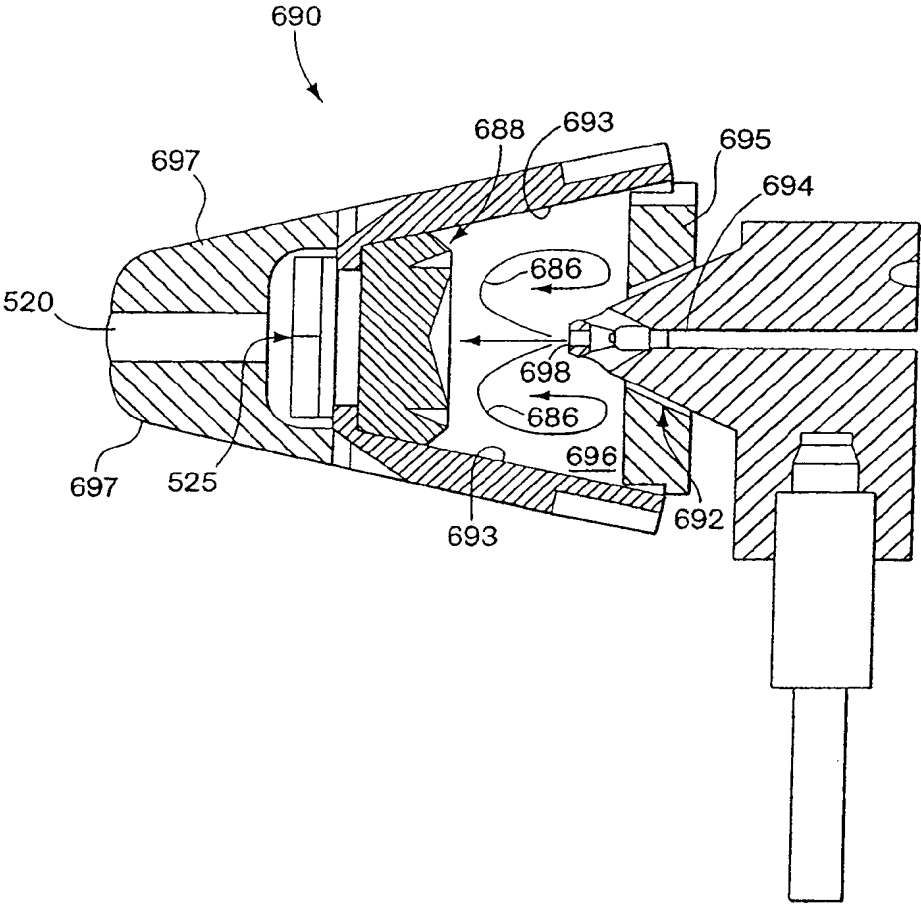


图 6D

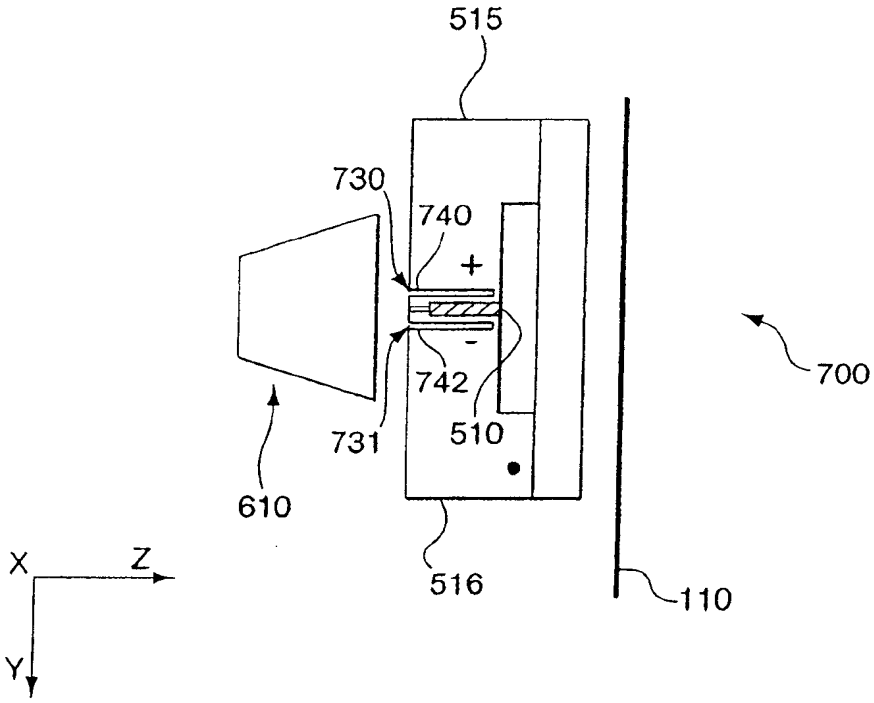


图 7

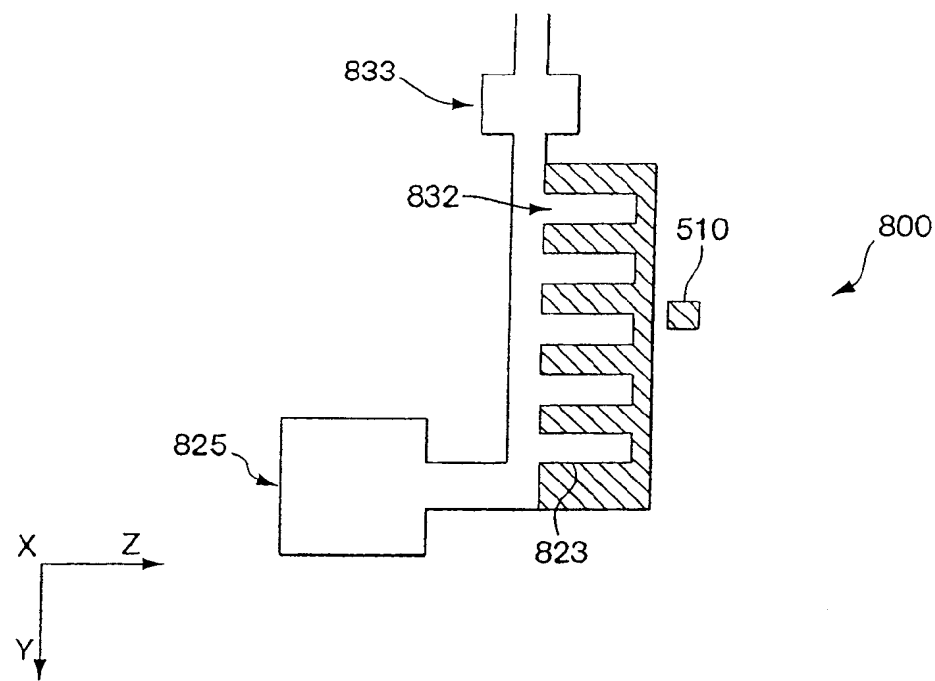


图 8

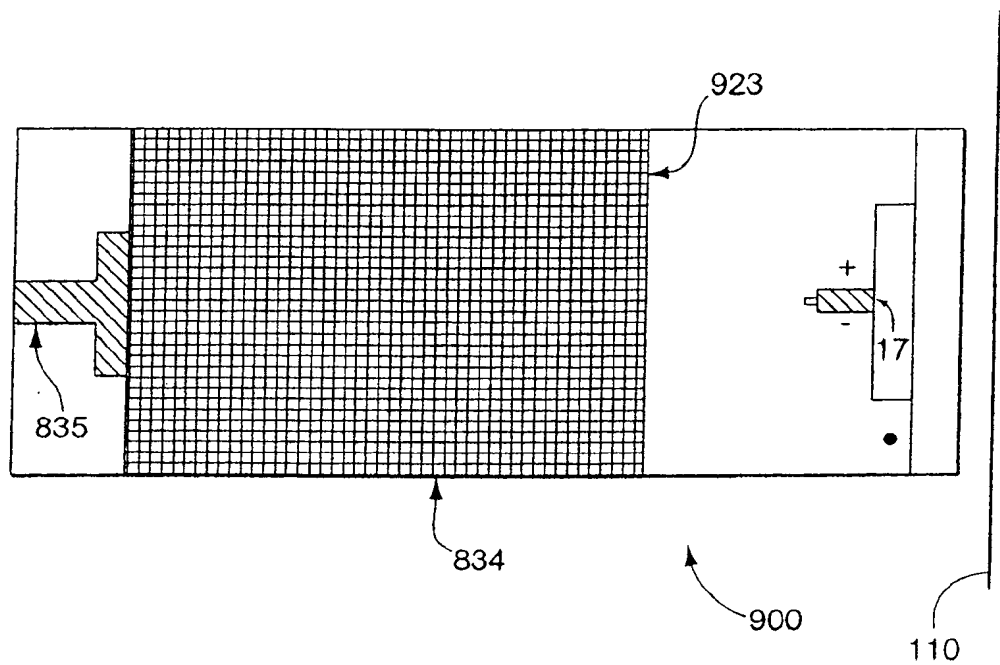


图 9

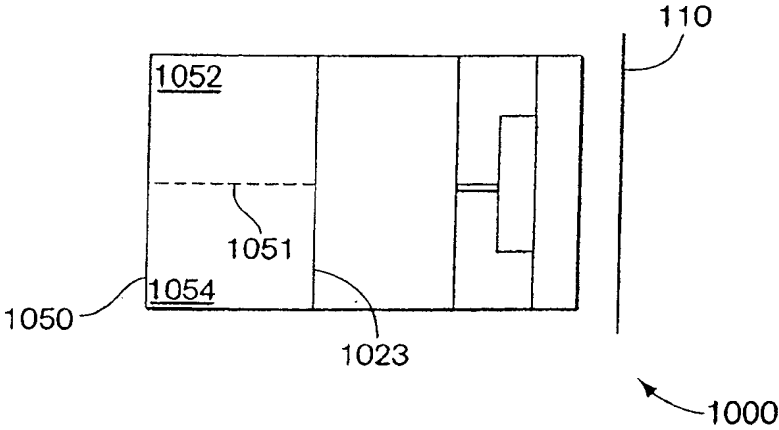


图 10

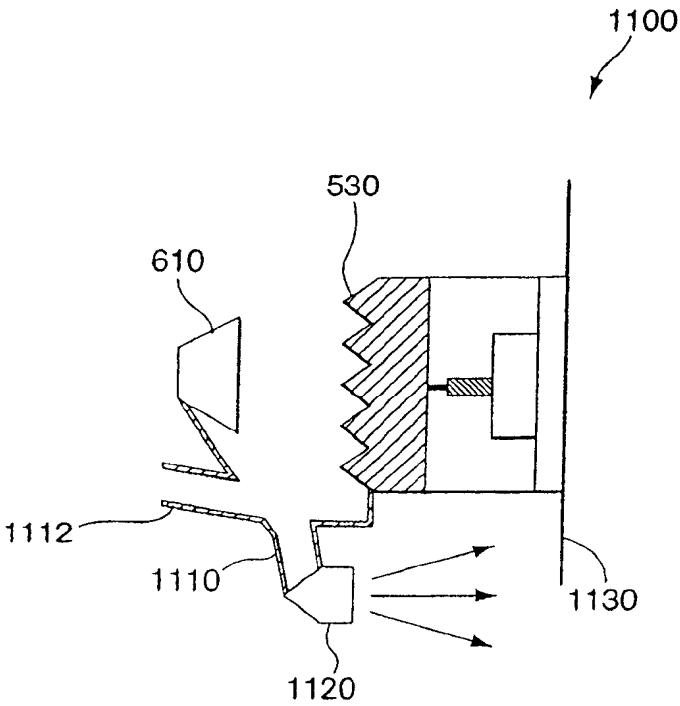


图 11

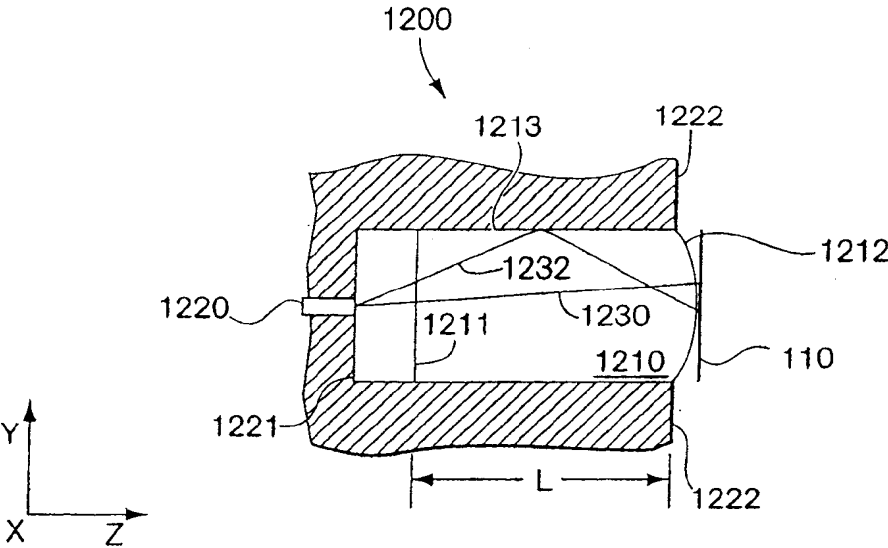


图 12A

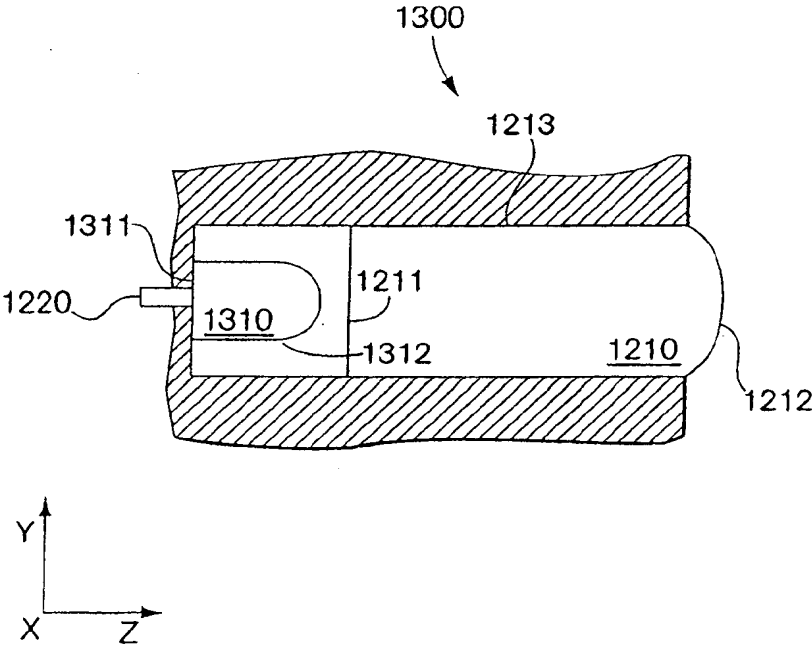


图 13A

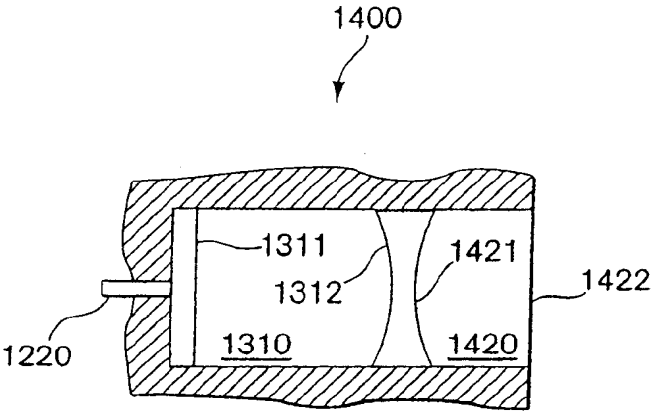


图 14A

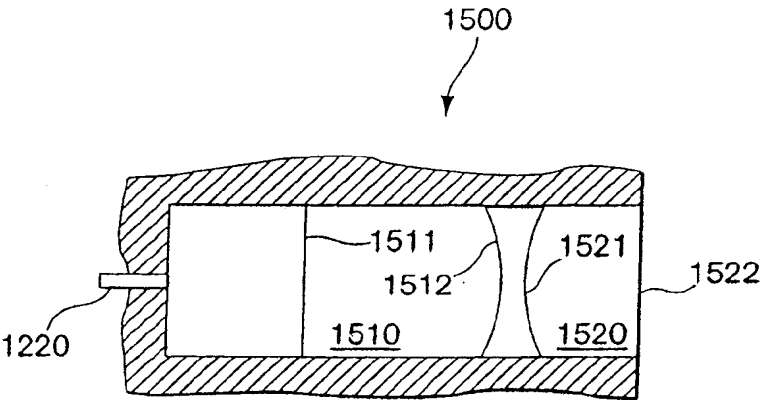


图 15A

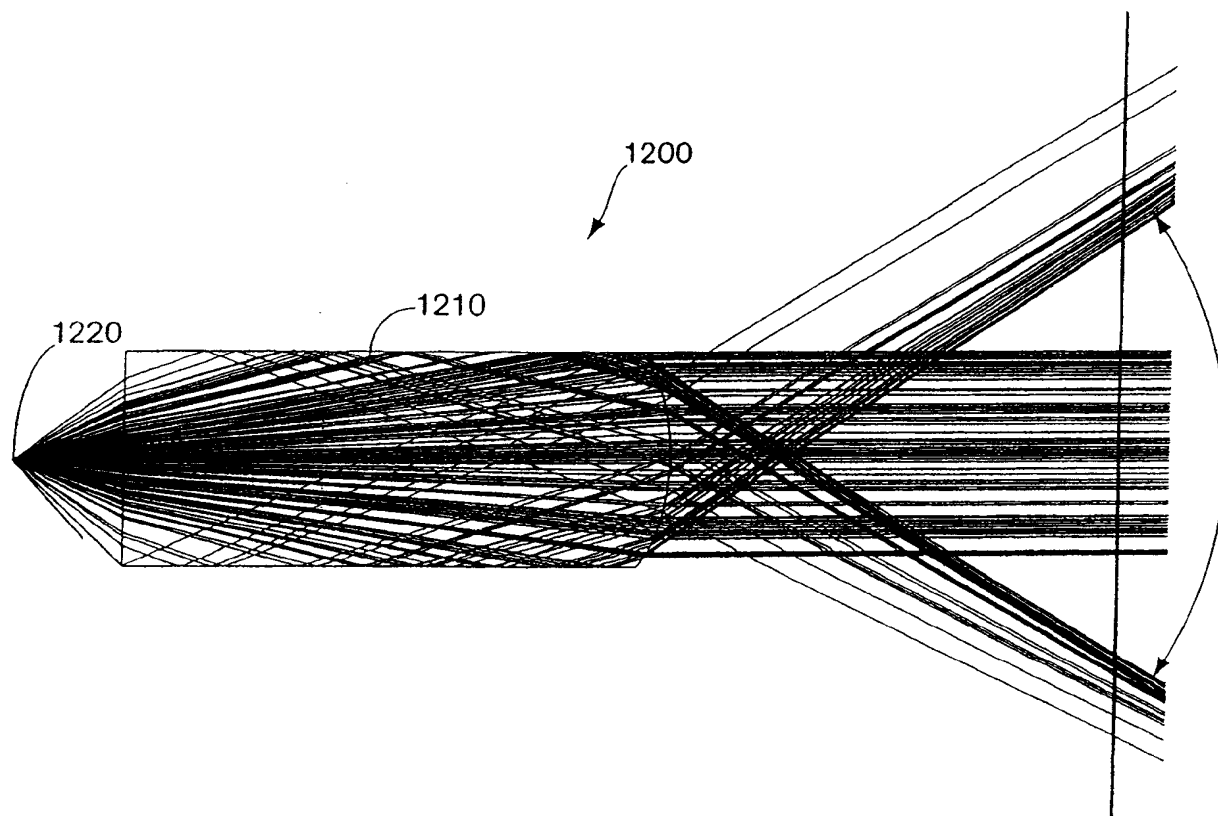


图 12B

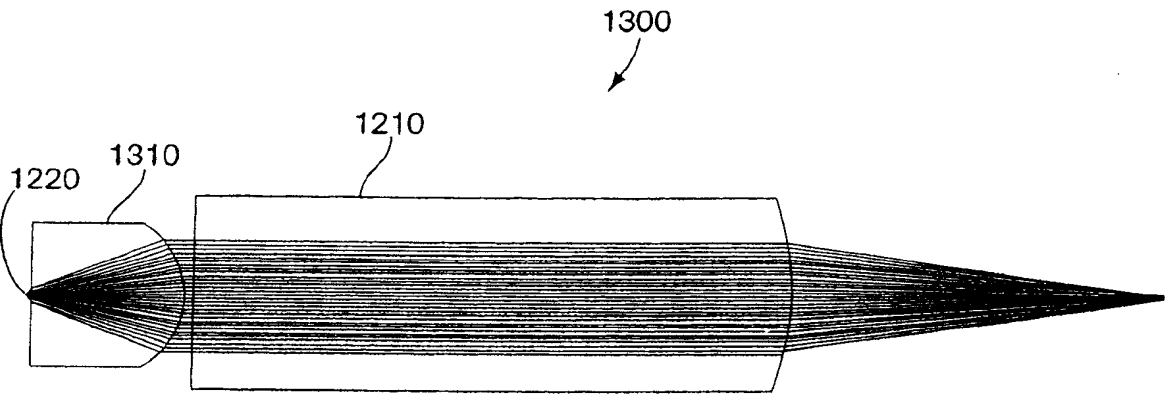


图 13B

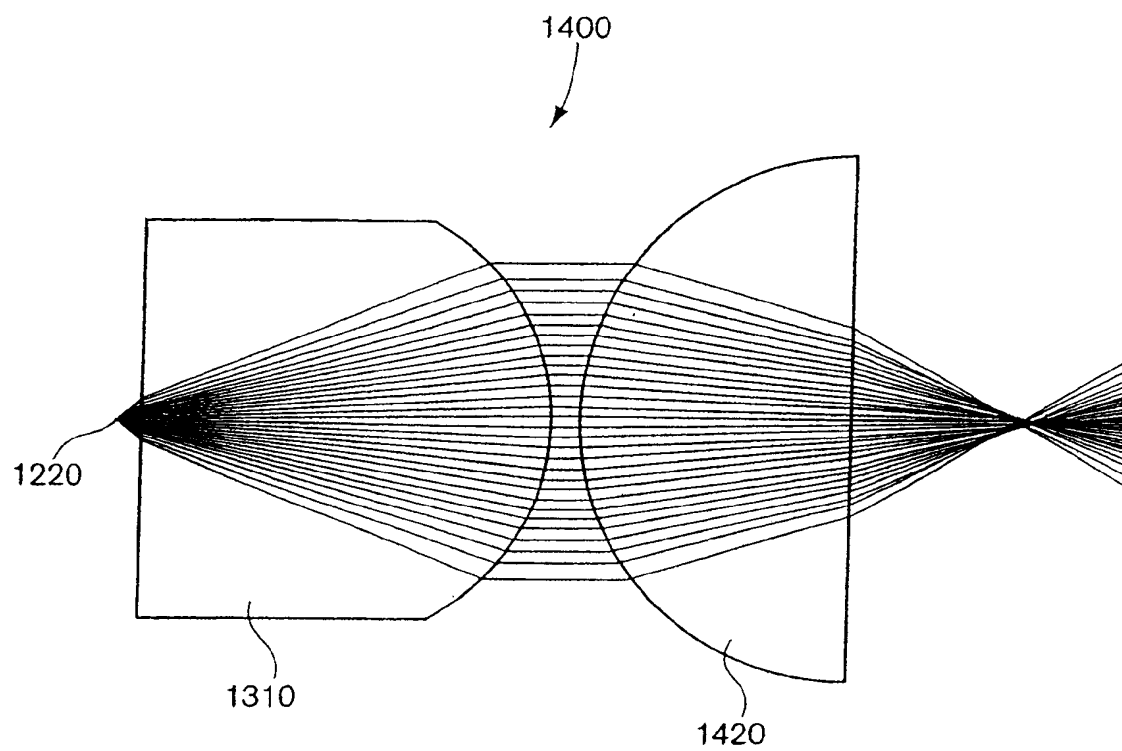


图 14B

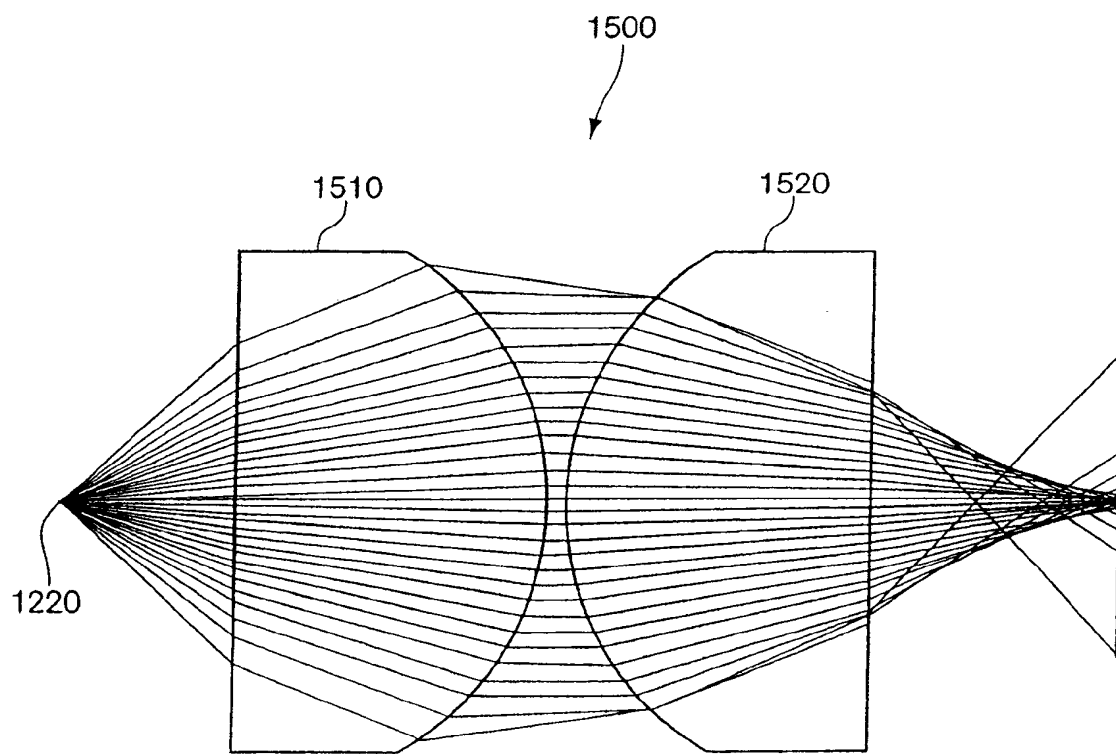


图 15B

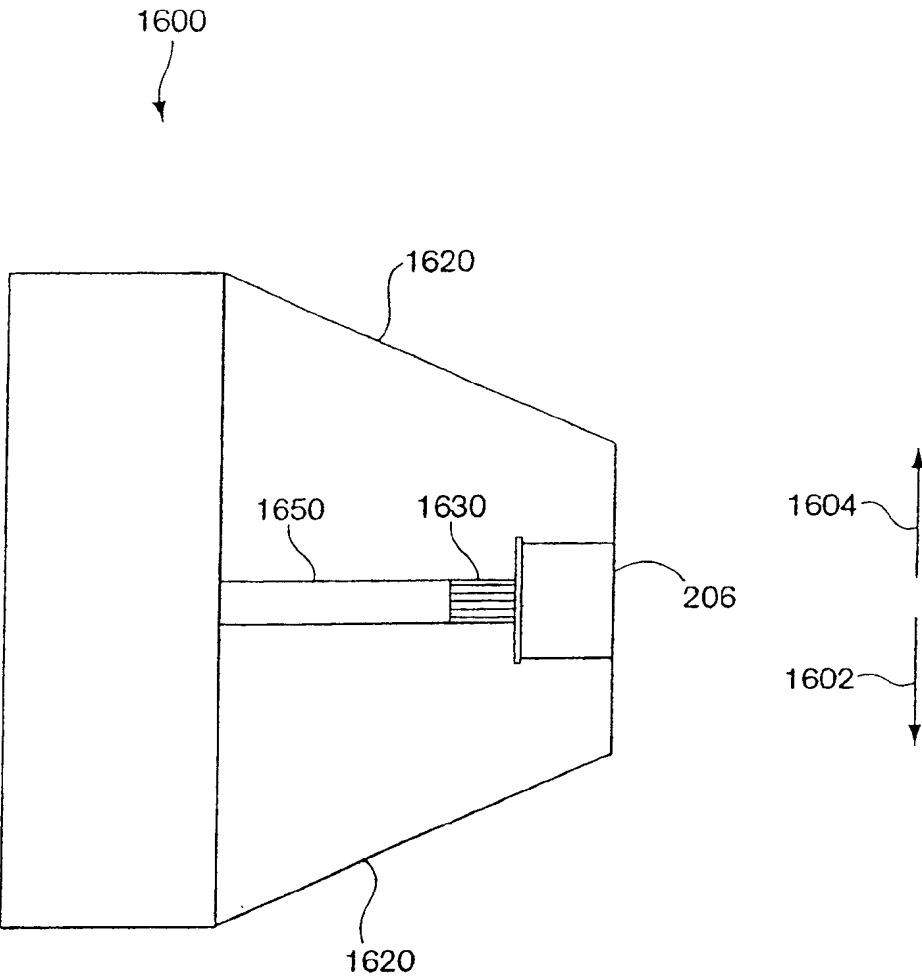


图 16A

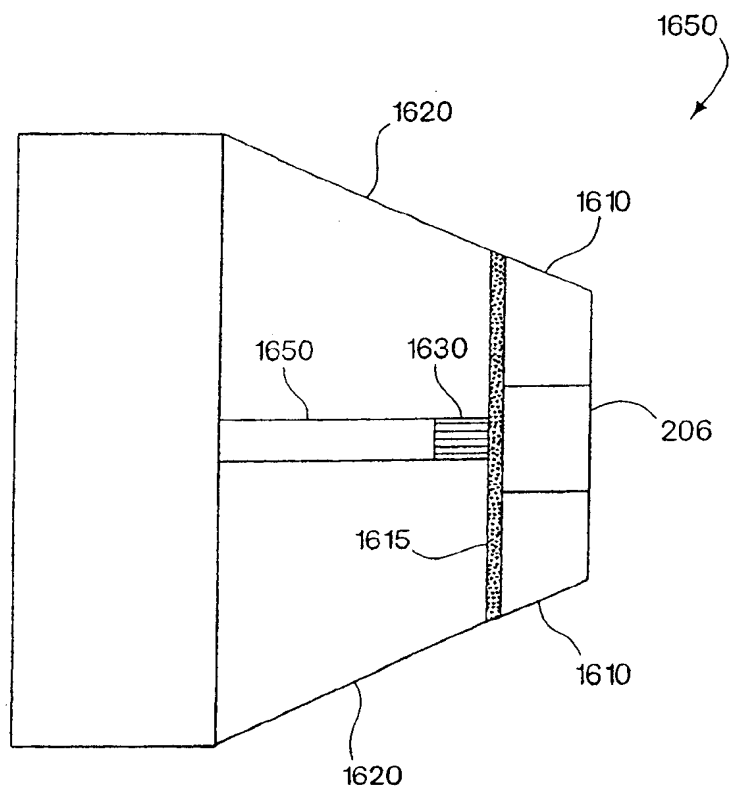


图 16B

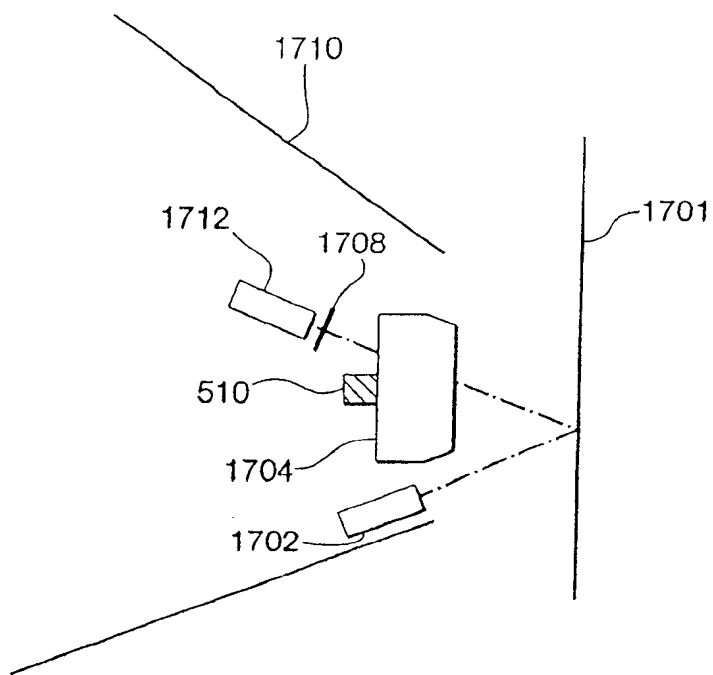


图 17A

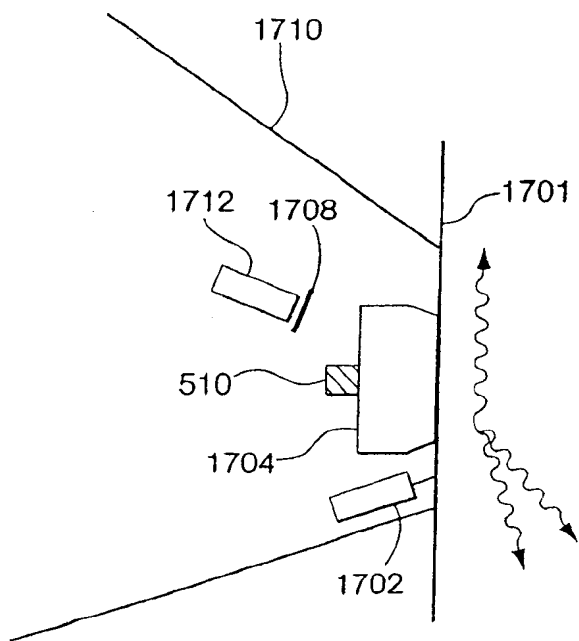


图 17B

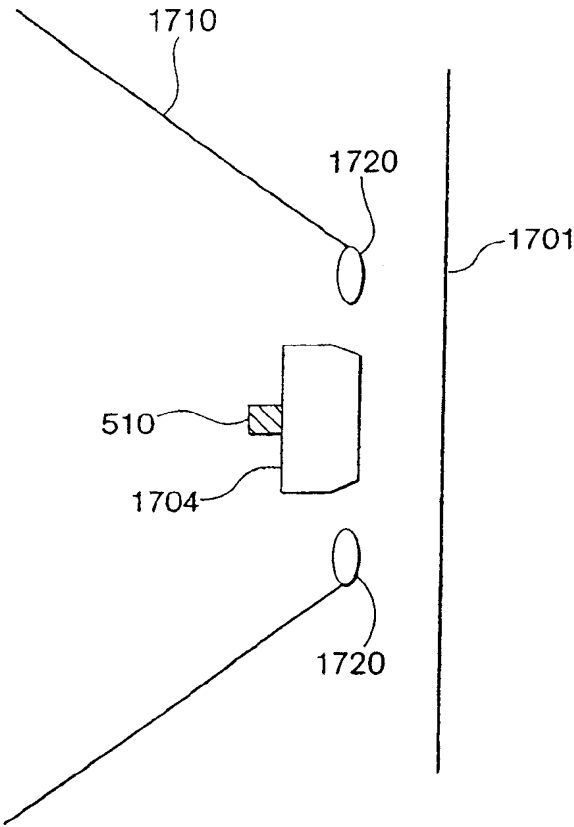


图 17C

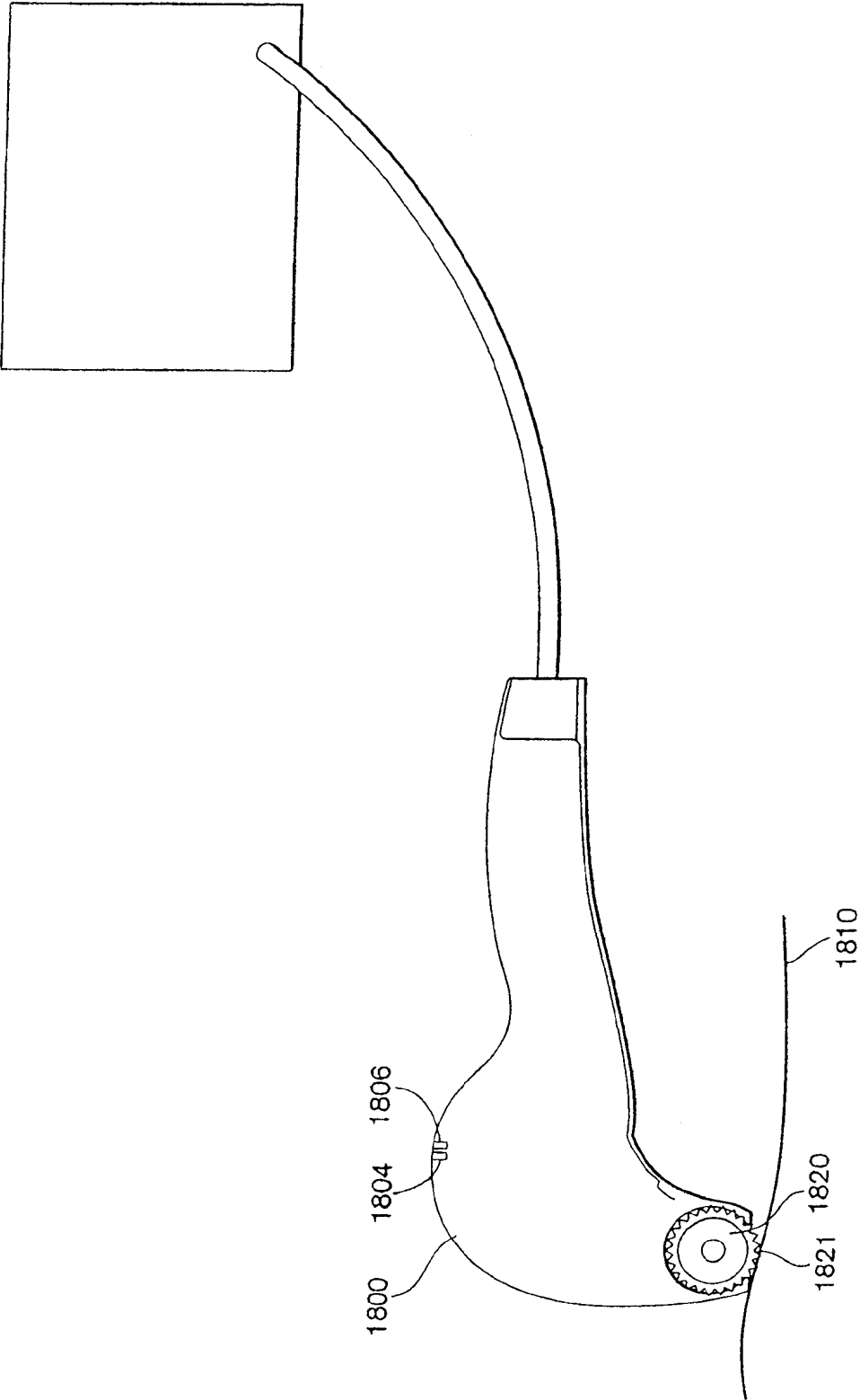


图 18A

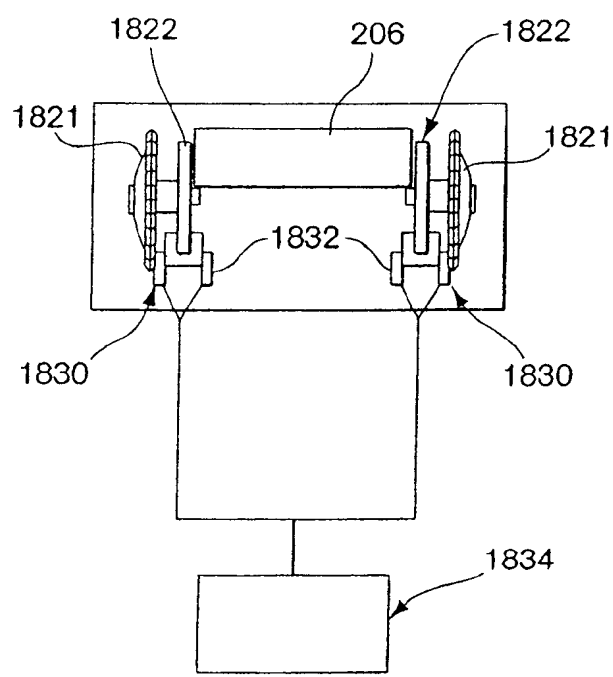


图 18B

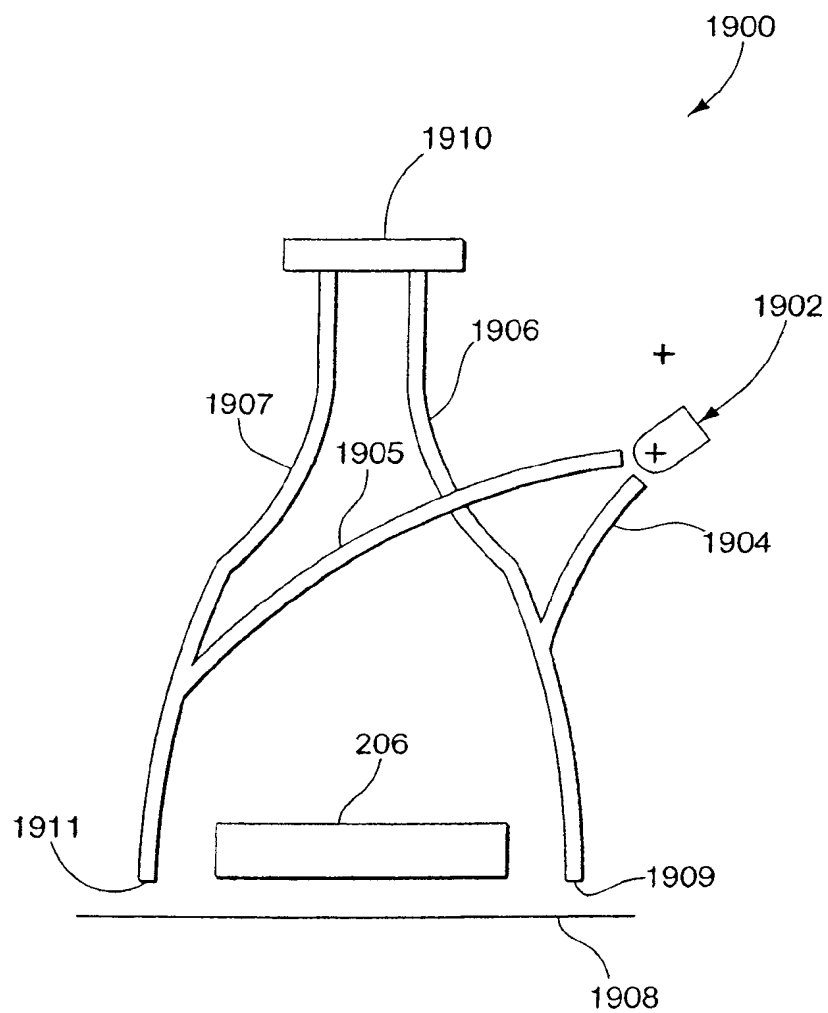


图 19

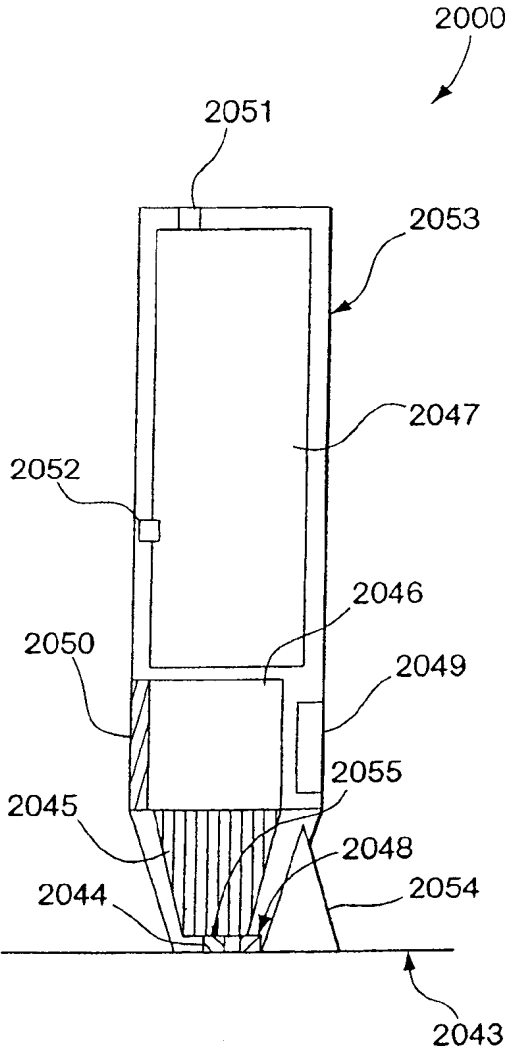


图 20

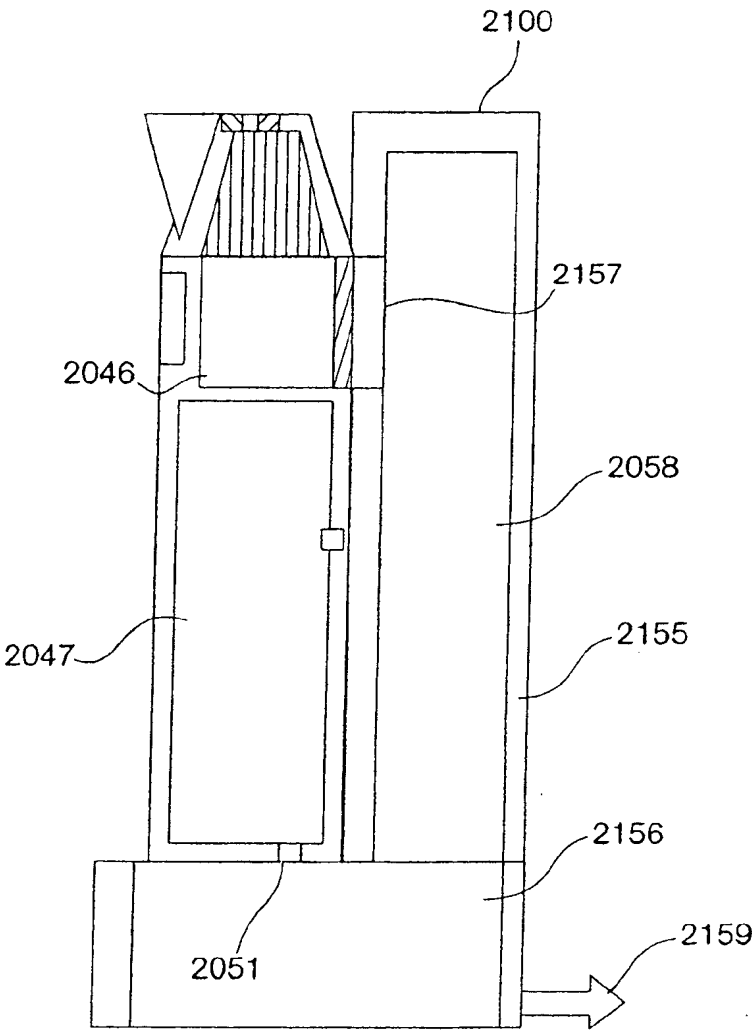


图 21

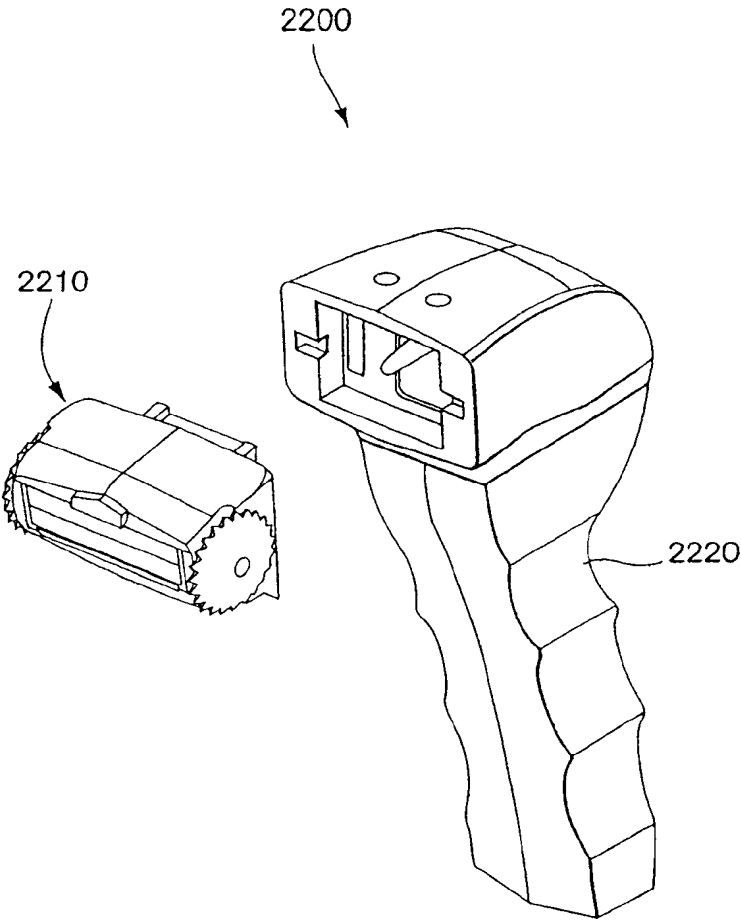


图 22

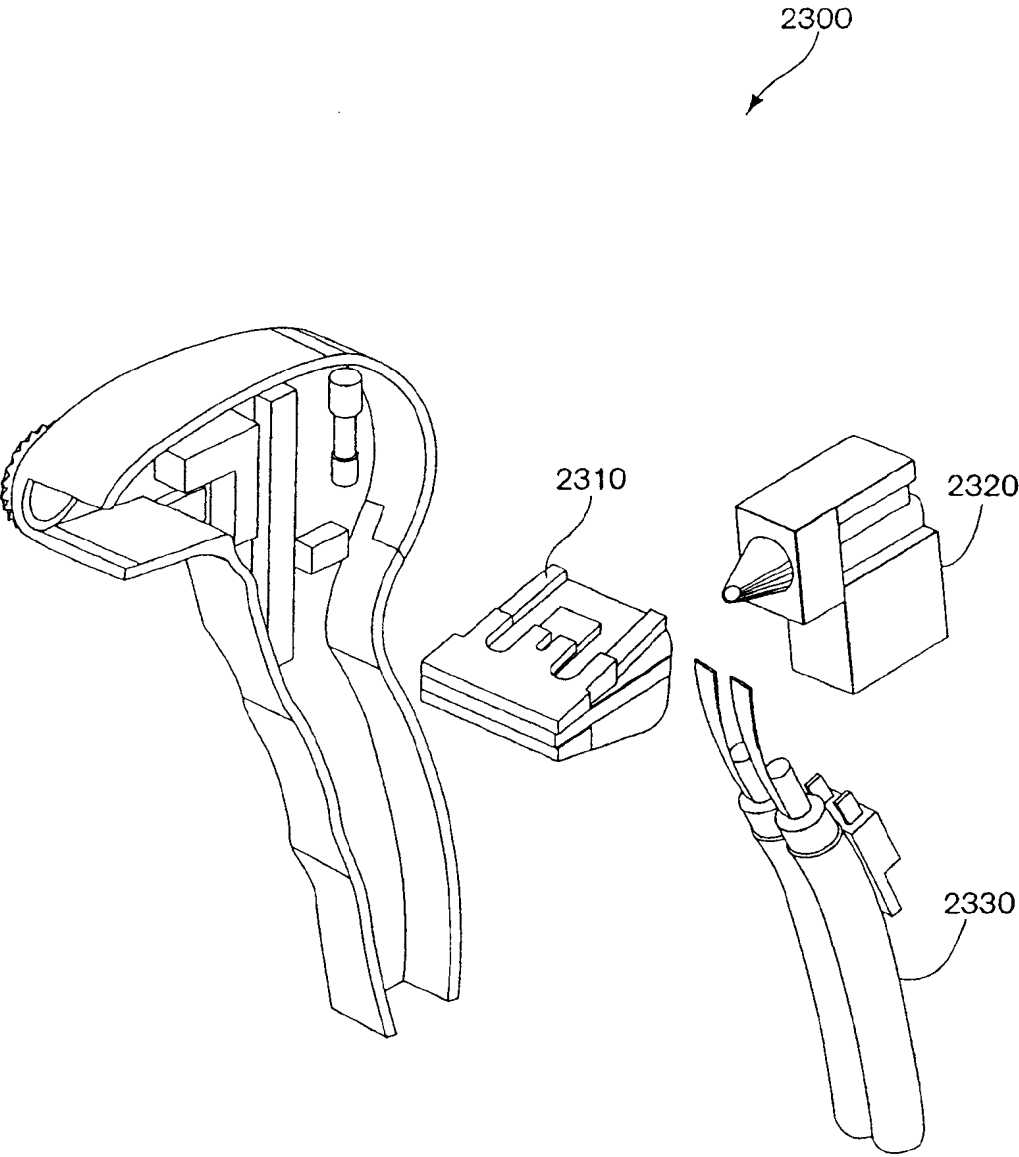


图 23

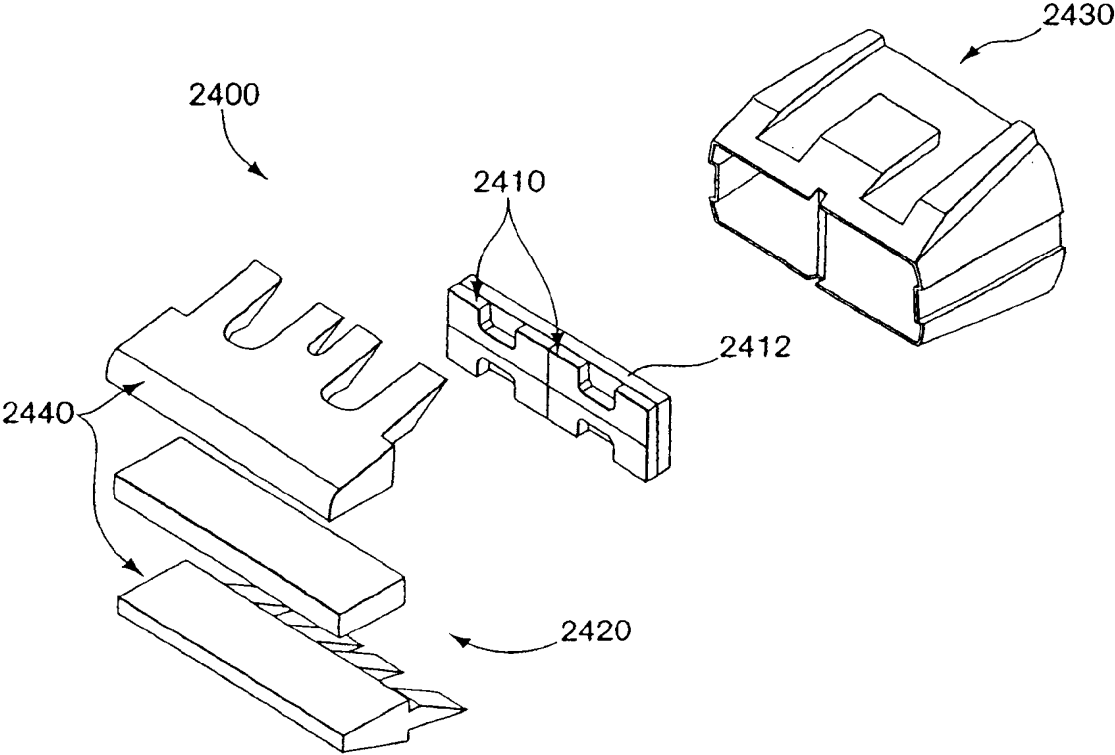


图 24

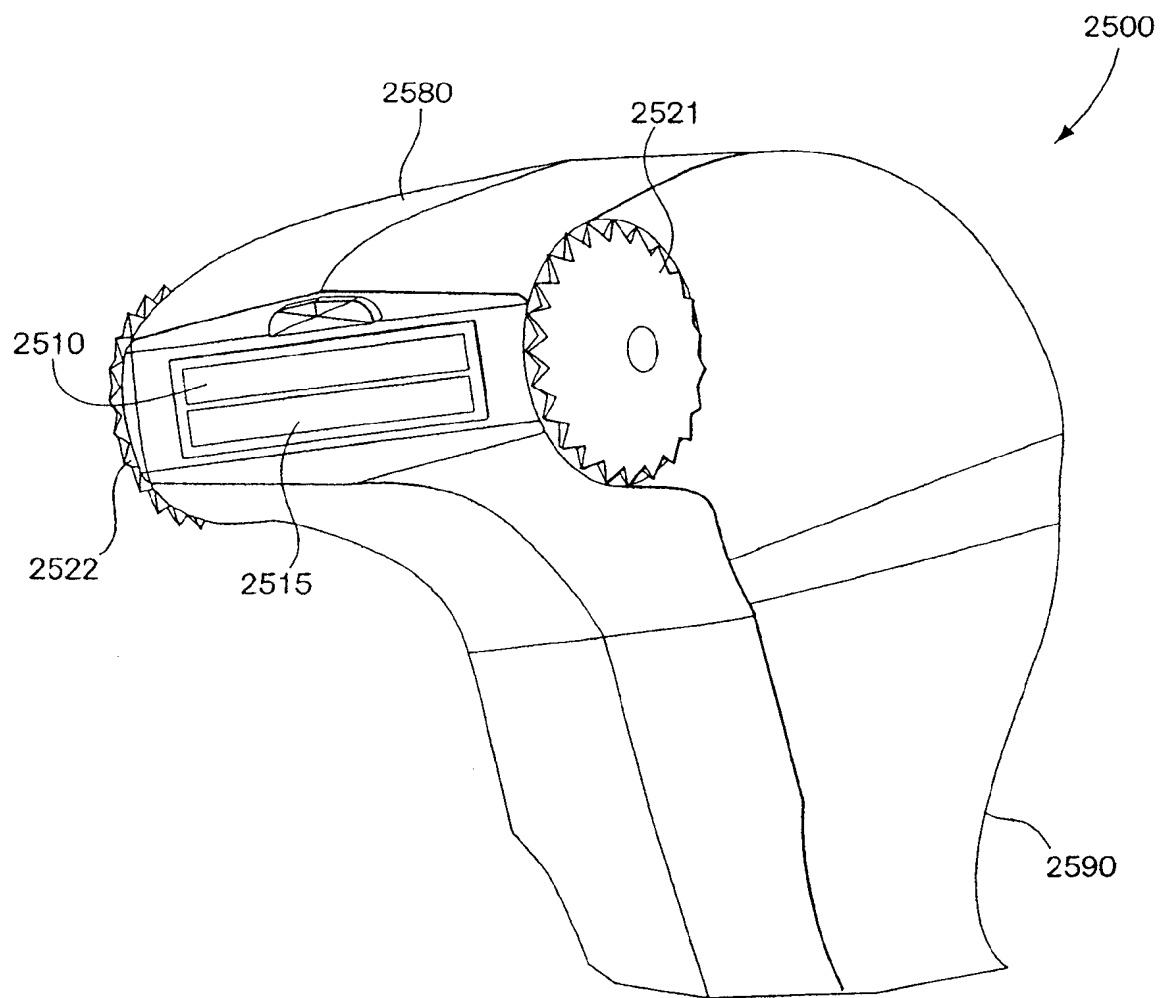


图 25

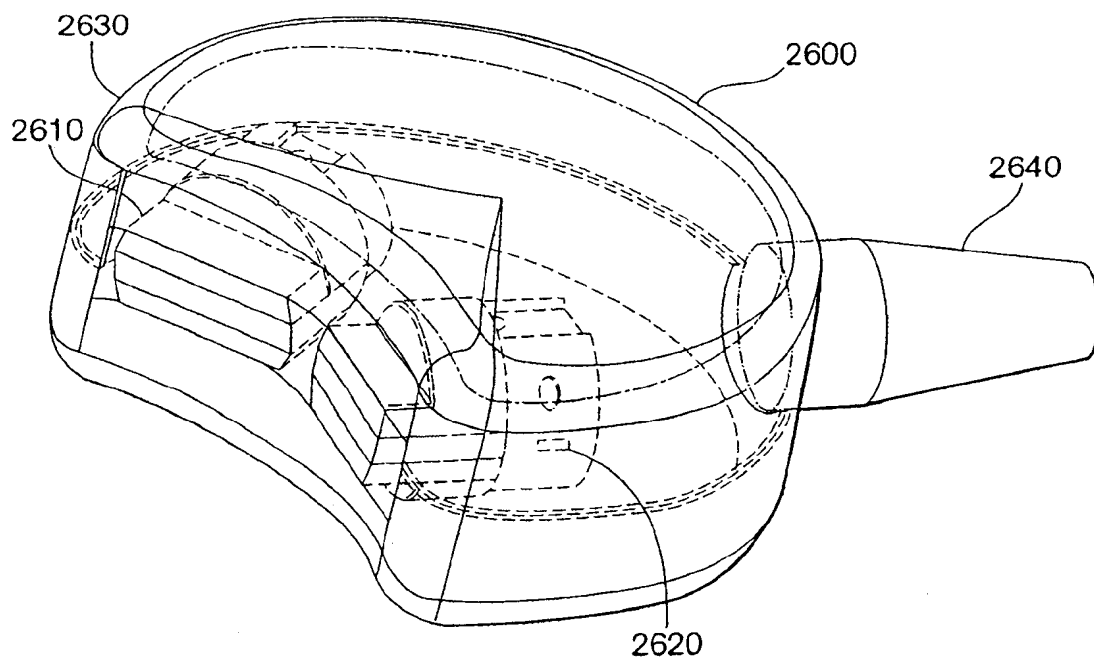


图 26A

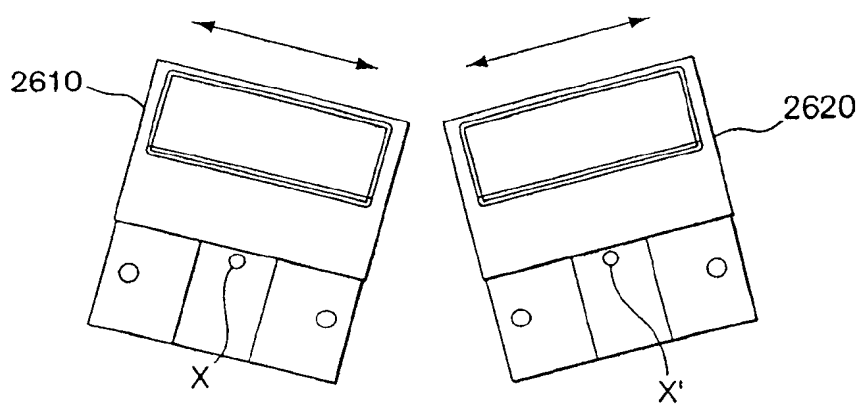


图 26B

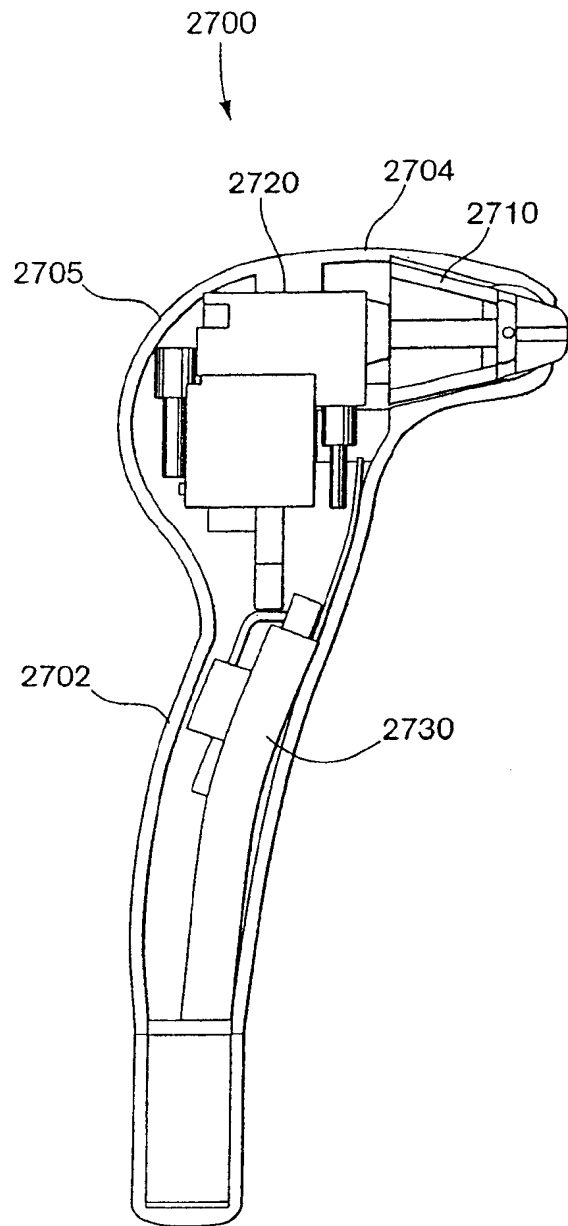


图 27

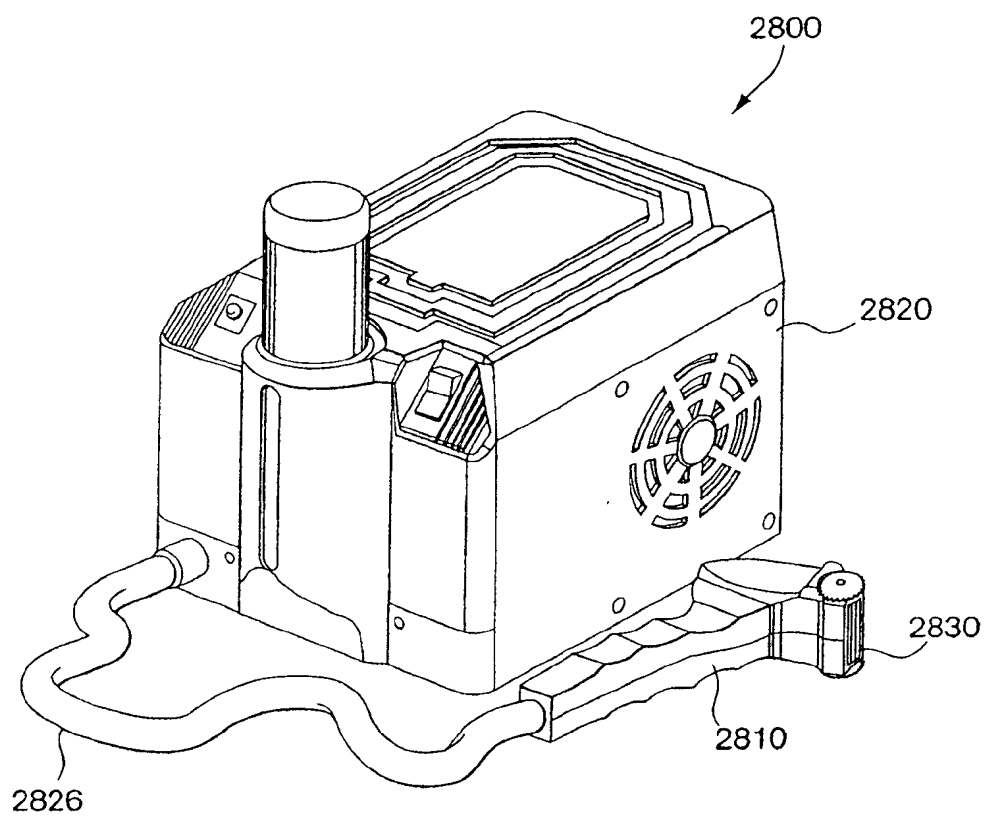


图 28

RADIATION TREATMENT APPARATUS**Publication number:** CN2053926 (U)**Publication date:** 1990-03-07**Inventor(s):** JIESHENG LI [CN]**Applicant(s):** LI JIESHENG [CN]**Classification:****- international:** **A61N5/06; A61N5/06;** (IPC1-7): A61N5/06**- European:****Application number:** CN19892006876U 19890506**Priority number(s):** CN19892006876U 19890506Abstract of **CN 2053926 (U)**

The utility model relates to an improved medical appliance, belonging to a TDP radiation treatment apparatus which is composed of a radiator (1) and a parabolic reflecting mirror (2), wherein, the radiator (1) is positioned at a focal point; as a result, radiation efficiency (comprising electromagnetic waves and 'particles') is enhanced. A frame-shaped base (4) is attached to a human body through a lacing (3) so that the problem of a support structure is conveniently solved. The relative positions of the reflecting mirror (2) and the radiator (1) can be adjusted so that radiation intensity is adjusted. Under the condition of unchanged work characteristics and medical characteristics, the cost of a TDP radiation treatment apparatus can be reduced to around 20 yuan, and power expenditure is tens of watts.

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[19] 中华人民共和国专利局

[11] 公告号 CN 2053926U



(12) 实用新型专利申请说明书

[21] 申请号 89206876.0

[51] Int.CI⁵
A61N 5/06

[43] 公告日 1990年3月7日

[22] 申请日 89.5.6

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[72] 设计人 李杰生

[39]

说明书页数: 3

附图页数: 2

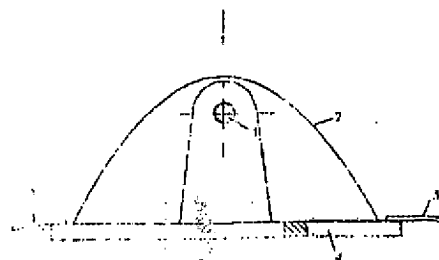
[43]

[54] 实用新型名称 新型辐射治疗器

[57] 摘要

本实用新型涉及一种改进的医疗器械——TDP 辐射治疗器, 由位于焦点处的辐射体(1)和抛物面反射镜(2)组成, 能大大提高辐射效率(包括电磁波与“微粒”); 由系带(3)将框形底座(4)缚于人体上, 十分方便地解决了支架问题。反射镜(2)和辐射体(1)的相对位置可调, 从而调节了辐射强度。在工作特性和医疗特性不变的条件下, 可使每台成本下降到二十元左右, 耗电数十瓦。

[39]



(BJ)第1452号

[43]

权 利 要 求 书

1. 一种辐射治疗器，其辐射体表面有TDP涂层，或者是与TDP涂层的成份基本相同的涂层，其特征在于：辐射体〔1〕置于抛物面反射镜〔7〕的焦点位置，辐射体尺寸相对抛物镜相当细小。

2. 一种辐射治疗器，其辐射体表面有TDP涂层，或者是与TDP涂层的成份基本相同的涂层，其特征在于：治疗器上联有系带〔9〕，系带有加长部份。

3. 权利要求1所述的辐射治疗器，其特征在于：以电热瓷芯作为发热元件，其外套以薄钢管，上面涂有TDP涂层；或者将TDP涂层直接涂在瓷芯上面。

4. 权利要求1所述的辐射治疗器，其特征在于：抛物镜〔7〕固定在外壳〔5〕上，外壳〔5〕上有散热孔〔10〕；在抛物镜〔7〕的两头，倾斜放置有两只反射镜〔8〕。

5. 权利要求〔2〕所述的辐射治疗器，其特征在于：有一中空的框形底座〔3〕，在其边沿上两对称位置连接系带〔9〕。系带〔9〕有加长部份。

6. 权利要求4和权利要求5所述的辐射治疗器，其特征在于：在底座〔3〕上面整有四根金属杆〔4〕，金属杆〔4〕插入外壳〔5〕内壁上固定的四只弹簧片〔6〕内。金属杆〔4〕顶端有限位凸起。

7. 权利要求4和权利要求5所述的辐射治疗器，其特征在于：辐

射体〔1〕通过支架〔2〕固定在底座〔3〕上面，当外壳〔5〕与底座〔3〕相对距离最小时，辐射体〔1〕正处于抛物镜〔7〕的焦点位置上。

说明书

新型辐射治疗器

本实用新型涉及一种改进的医疗器械——TDP辐射治疗器或TDP类辐射治疗器。这里说的TDP类，即辐射体涂料成份和工作温度都与TDP基本相同。

TDP辐射治疗器辐射体的工作温度高达数百度。有相当面积的辐射体在这样高的温度下，不能近距离照射人体，一般距离在30cm左右。TDP辐射器的辐照度大致按距离的平方倍衰减。因此人体承受的有效辐射很少。这样，一方面耗能大，能量利用率很低；另一方面需要一个笨重、昂贵的支架。

以上现有技术在中国专利CN85100830A和沈阳军区总医院金石正、张衍平著《对特定电磁波治疗器某些物理特性的研究》(载《特定电磁波研讨会论文选编》7~12页)，以及各TDP厂家的使用说明书中均有反映。

本实用新型在结构上作了重大改进，大幅度降低了TDP辐射治疗器的能耗，取消了支架结构。

本实用新型将TDP辐射体的尺寸收缩得很小，使其置于一抛物面反射镜的焦点上，使电磁波成平行束向外辐射。将抛物镜开口扣在需要照射的部位上，并用系带将治疗器缚在机体上。这样一方面

辐射体本身可以保持较高的温度，保证了TDP涂层的辐射性能；另一方面通过抛物镜的反射，辐射能量能够比较均匀、柔和地照向人体，距离可以隔得很近，而且反射镜又将辐射定向射到一个方向，耗散很小。由于辐射体与人体的距离大大缩短，“微粒”辐射作用也大大加强，保证了这一方面的治疗作用。

下面结合附图，进一步介绍其结构。图1是主剖视图，图2是右剖视图，图3是仰视图。本图例中的细管形辐射体〔1〕，由电热瓷芯，在外面套上涂有TDP涂层的薄钢管，或将涂层直接涂在瓷芯外表面而构成。辐射体〔1〕通过支架〔2〕固定在中空的框形底座〔3〕上面。底座〔3〕上面整有四根金属杆〔4〕，金属杆〔4〕顶端有限位凸起。金属杆〔4〕插入外壳〔5〕内壁上固定的四只弹簧片〔6〕内。外壳〔5〕在外力作用下，可沿金属杆〔4〕上下移动，以调节与底座〔3〕的相对距离，而又不会与底座脱离。外壳〔5〕内安装有一个抛物面反射镜〔7〕。在抛物镜〔7〕的两头另外倾斜安装有两只反射镜〔8〕。底座〔3〕两侧对称位置上安有两根彼此可以扣在一起的系带〔9〕，另有备用带可以将系带〔9〕接长，以满足不同部位的需要。用系带〔9〕可将底座〔3〕连同整个治疗器缚在人体需要照射的部位上面。当外壳〔5〕滑动到最低位置时，即它与底座〔3〕的距离最近时，支架〔2〕保证辐射体〔1〕位于抛物镜〔7〕的焦点位置。此时机体受到辐射最强，将外壳〔5〕向上滑动，则辐射体〔1〕离开抛物镜〔7〕

的焦点位置，机体受到的辐射减弱。由此调节辐射强度。外壳〔5〕侧壁上可适当开孔〔10〕以散热。

本实用新型能将一般TDP治疗器的耗电从三百瓦左右降低到数十瓦级，成本降到数十元。预热由十余分钟缩短到一、两分钟，壳体温度也接近人体，使用十分方便。安全，系上治疗器以后，人体仍可自由移动，并可加盖衣物、被盖，有助于患者的保暖和身体隐蔽。

采用20 W左右的电热瓷芯，将TDP材料直接涂在瓷芯表面，能实现最好的技术——经济效果，每台成本仅20元左右。

说明书附图

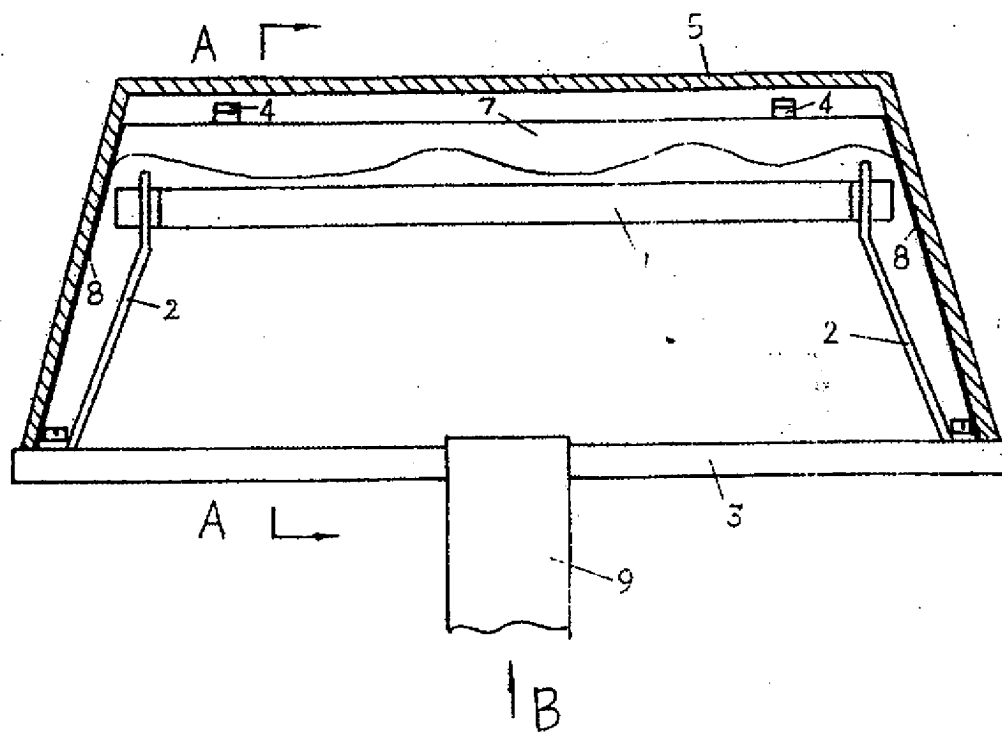


图 1

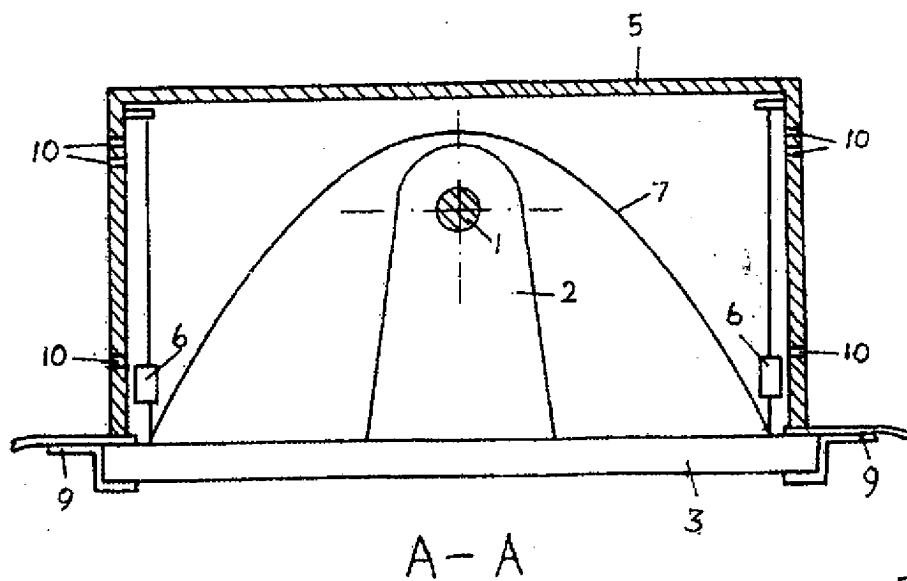


图 2

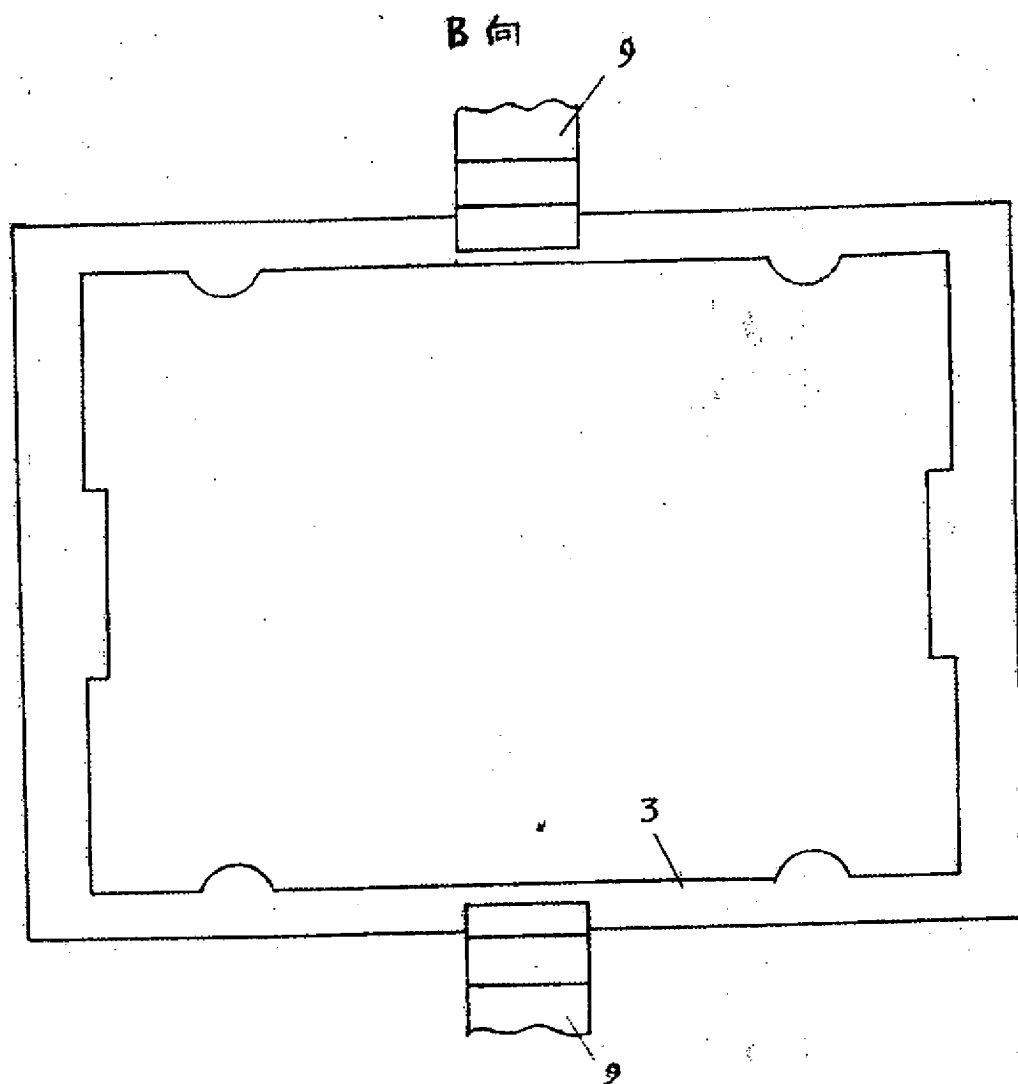





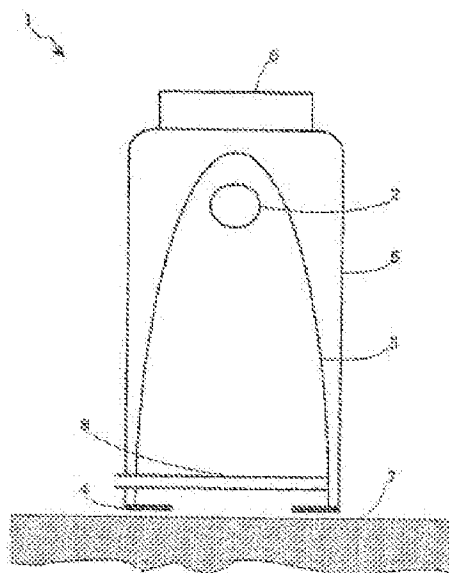


图 3

Irradiating device used for treating acne comprises a radiation source emitting a broad band spectrum in a specified region and operating in the pulse manner**Publication number:** DE10112289 (A1)**Publication date:** 2002-09-26**Inventor(s):** WILKENS JAN HENRICK [DE]**Applicant(s):** OPTOMED OPTOMEDICAL SYSTEMS GM [DE]**Classification:****- international:** **A61N5/06**; A61B18/18; **A61N5/06**; A61B18/18; (IPC1-7): A61N5/06**- European:** A61N5/06C2**Application number:** DE20011012289 20010308**Priority number(s):** DE20011012289 20010308**Cited documents:** DE19524461 (A1) DE4143168 (A1) DE29613075U (U1) DE9321497U (U1) US6183500 (B1)[more >>](#)**Abstract of DE 10112289 (A1)**

Irradiating device comprises a radiation source (2) emitting a broad band spectrum in the region of 320-670 nm and operating in the pulse manner with an energy of less than 3 J/cm² and a power density of less than 2 W/cm². An Independent claim is also included for a process for the treatment of acne using the above irradiating device. Preferred Features: A luminescent material film (8) is arranged in front of the radiation source and consists of a silicone elastomer doped with inorganic luminescent material particles, e.g. Sr₂P₂O₇ fluorescing in the spectral region of 410-490 nm.



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①9 BUNDESREPUBLIK
DEUTSCHLAND



DEUTSCHES
PATENT- UND
MARKENAMT

⑫ **Offenlegungsschrift**
⑩ **DE 101 12 289 A 1**

⑤1 Int. Cl.⁷:
A 61 N 5/06

⑳ Aktenzeichen: 101 12 289.6
㉔ Anmeldetag: 8. 3. 2001
㉓ Offenlegungstag: 26. 9. 2002

DE 101 12 289 A 1

㉑ Anmelder:

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㉒ Vertreter:

Patentanwälte Effert, Bressel und Kollegen, 12489
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㉑ Erfinder:

Wilkens, Jan Henrick, Dr., 66424 Homburg, DE

㉒ Entgegenhaltungen:

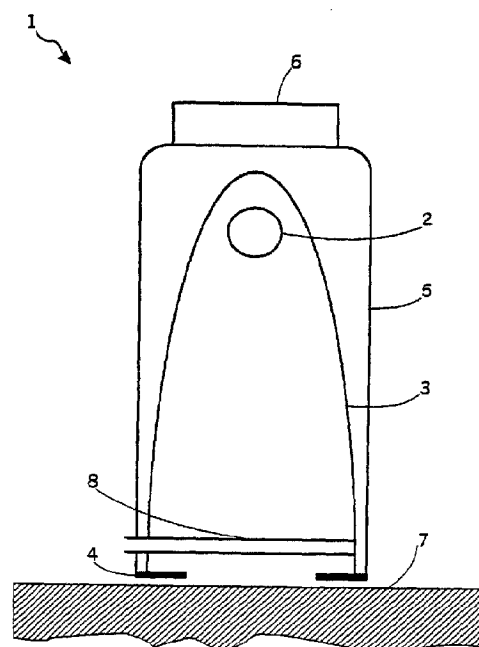
DE	195 24 461 A1
DE	41 43 168 A1
DE	296 13 075 U1
DE	93 21 497 U1
US	61 83 500 B1
EP	07 26 083 A2
EP	05 92 794 A2
WO	00 28 575 A1
WO	00 02 491 A1

Die folgenden Angaben sind den vom Anmelder eingereichten Unterlagen entnommen

Prüfungsantrag gem. § 44 PatG ist gestellt

㉒ Bestrahlungsanordnung und Verfahren zur Behandlung von Akne

㉒ Die Erfindung betrifft eine Bestrahlungsanordnung (1) und ein Verfahren zur Behandlung von Akne, umfassend mindestens eine Bestrahlungsquelle (2), wobei die Bestrahlungsquelle (2) mindestens ein breitbandiges optisches Spektrum im Bereich von 320 bis mindestens 670 nm emittiert, die Bestrahlungsquelle (2) im Pulsbetrieb betreibbar ist und die Pulsenergie kleiner 3 J/cm^2 und die mittlere Leistungsdichte kleiner 2 W/cm^2 ist.



DE 101 12 289 A 1

[0001] Die Erfindung betrifft eine Bestrahlungsanordnung und ein Verfahren zur Behandlung von Akne.

[0002] Es ist bekannt, Akne, eine aufgrund von Bakterienwachstum in verstopften Follikeln talgdrüsenreicher Hautbezirke mit Verhornungsstörungen hervorgerufene Hauterkrankung, mit blauem Licht im Bereich von 400–440 nm ohne wesentliche UVA-Anteile zu behandeln, wobei die Erfolge beschränkt blieben. Hierzu sei auf den Fachartikel "V. Sigurdsson et al. Phototherapy of Acne Vulgaris with visible Light, *Dermatologie* **1997; 194**; Bd. 3, 256–260" mit weiteren Literaturhinweisen verwiesen. Angestoßen wurde diese Form der Therapie, daß Aknefollikel im Rahmen der dermatologischen Untersuchung mit einer sogenannten "wood-lamp" rot fluoreszieren. Als Quelle der Fluoreszenz wurde die Speicherung großer Mengen von Porphyrinen im Propionibakterium acne nachgewiesen. (Mc Ginley et al., *Facial follicular porphyrin fluorescence. Correlation with age and density of propionibacterium acnes*, Br. J. Dermatol Vol. 102., Bd. 3, 437–441, 1980). Da Porphyrine ihre Hauptabsorption (Soret-band) um 420 nm haben, war es für Melfert et al. naheliegend, bakterielle Aknefollikel mit blauem Licht zu behandeln. Die langwelligste Absorptionsbande der Porphyrine liegt bei 630 nm mit einer Eindringtiefe von 4 mm, die für eine photodynamische Follikelbehandlung am besten geeignet ist und auch verwendet wird.

[0003] Aus der WO 00/02491 ist eine derartige Bestrahlungsanordnung bekannt, die mindestens ein schmalbandiges Spektrum im Bereich von 405–440 nm umfaßt. Als alternative oder kumulative Spektralbereiche sind die Wellenlängenintervalle von 630–670 nm bzw. 520–550 nm angegeben. Zur weiteren Verbesserung des Wirkungsgrades wird vorgeschlagen, die zu bestrahlende Partie mit Sauerstoff anzureichern, indem mit Sauerstoff angereicherte Emulsionen vor oder während der Bestrahlung auf die zu bestrahlende Fläche aufgetragen werden. Die Bestrahlungsstärke liegt dabei zwischen 10–500 mW/cm².

[0004] Weiter ist aus der WO 00/64537 eine Bestrahlungsanordnung zur Behandlung von Akne bekannt. Dabei wird die zu behandelnde Fläche mit UV-Licht im Bereich von 320–350 nm behandelt. Die dabei eingestrahelte Energie wird mit 1–5 J/cm² angegeben. Bei Verwendung eines Lasers soll dabei die Pulsenergie zwischen 5–25 mJ/cm² liegen, so daß sich bei Pulslängen von 10 ns Bestrahlungsstärken von ca. 2 MW/cm² einstellen. Die bekannte Bestrahlungsanordnung geht dabei von der Erkenntnis aus, daß sonnenähnliche Spektren nicht zur Behandlung von Akne geeignet sind, sondern vielmehr sogar Akneschübe auslösen können.

[0005] Aus der EP 0 565 331 B1 ist eine Vorrichtung zur Behandlung von Gefäßerkrankungen in einem Bereich der Haut bekannt, umfassend ein Gehäuse, mit einer inkohärenten Lichtquelle, montiert in dem Gehäuse und geeignet zum Produzieren von gepulstem Licht für die Behandlung und eine Öffnung in dem Gehäuse, welche einen austretenden Lichtstrahl bestimmt, der auf den Hautbehandlungsbereich gesendet wird, ohne durch ein Kabel aus optischen Fasern zu gehen, und der einen breiteren Strahlungsbereich aufweist als Vorrichtungen mit Kabel aus optischen Fasern, wobei die Vorrichtung ein die niedrigen Frequenzen abschneidendes Filter umfaßt, um die sichtbaren und ultravioletten Teile des Spektrums herauszuschneiden und die inkohärente Lichtquelle einen Ausgangslichtstrahl mit Wellenlängen im Bereich zwischen 300 und 1000 nm produziert. Die Lichtquelle ist elektrisch mit einer Variabel-Impulsbreiten-Erzeugerschaltung verbunden, um einen geregelten Zeitimpuls zu liefern mit einer Breite zwischen 1 und 10 ms, wobei der

austretende Lichtstrahl auf der Haut eine Energiedichte zwischen 30 und 100 J/cm² erzeugt, so daß der hinaustretende Lichtstrahl nach Durchgang durch das obengenannte, die niedrigen Frequenzen abschneidende Filter in die Haut so tief wie gewünscht eindringen kann, ohne die Haut zu verbrennen, um ein unter der Haut und innerhalb des Hautbehandlungsbereiches liegendes Blutgefäß zu erwärmen und im Blutgefäß Blutkoagulation zu verursachen. Die dort beschriebene Blutkoagulation ist bei der Behandlung von Akne zu vermeiden, so daß die dort beschriebene Vorrichtung zur Behandlung von Akne oder anderer oberflächlicher Hauterkrankungen ungeeignet ist.

[0006] Die bekannten Bestrahlungsanordnungen zur Behandlung von Akne sind jeweils technisch sehr aufwendig und damit kostenintensiv, insbesondere wenn in den ausgewählten Spektralbereichen die hohen Energiedichten gefordert werden.

[0007] Der Erfindung liegt daher das technische Problem zugrunde, eine Bestrahlungsanordnung und ein Verfahren zur Behandlung von Akne zu schaffen, die kostengünstig realisierbar sind und einen hohen Wirkungsgrad aufweisen.

[0008] Die Lösung des technischen Problems ergibt sich durch die Gegenstände mit den Merkmalen der Patentansprüche 1, 12 und 13. Weitere vorteilhafte Ausgestaltungen der Erfindung ergeben sich aus den Unteransprüchen.

[0009] Hierzu ist die Bestrahlungsquelle als breitbandige Bestrahlungsquelle mit einem Wellenlängenbereich von mindestens 320 bis mindestens 670 nm ausgebildet, die im Pulsbetrieb betreibbar ist, wobei die Pulsenergie kleiner 3 J/cm² und die mittlere Leistung kleiner als 2 W/cm² beträgt. Unter mindestens 320 nm bedeutet dabei, daß die Bestrahlungsquelle durchaus auch kleinere Wellenlängen erzeugen kann, diese aber nicht auf die zu behandelnde Fläche weitergeleitet werden, sondern vorher unterdrückt werden. Erfindungsgemäß wird ausgenutzt, daß entgegen Ausführungen in der Literatur, durch den Pulsbetrieb die Bildung von Singulett-Sauerstoff um Größenordnungen gegenüber CW-Betrieb bei gleicher Energie erhöht wird. Da entgegen den Ausführungen in der WO 00/64537 auch sichtbare Spektralbereiche Behandlungswirksam sind, kann auf preiswerte breitbandige Bestrahlungsquellen zurückgegriffen werden, so daß kostenintensive Laser oder Filtermaßnahmen entbehrlich sind. Die Ursache für die Akneschübe mit solarähnlichen Strahlungsquellen liegt nämlich vermutlich nicht im sichtbaren Anteil, sondern in dem UVB-Anteil bis 320 nm, auf den erfindungsgemäß auch nicht zurückgegriffen wird. Ein weiterer Vorteil gegenüber der WO 00/02491 ist, daß eine größere Leistung bzw. Energie des blauen Spektralteils zum Follikel gelangt. Da die Follikel relativ tief unter der Haut sitzen, erreicht normalerweise nur ein Bruchteil des blauen Spektralbereichs zwischen 400–500 nm die Follikel, so daß Bestrahlungsanordnungen mit einem reinen Blauspektrum mit relativ hoher Leistung arbeiten müssen. Dies liegt zum einen an der gerigen Eindringtiefe des blauen Spektralanteils und zum anderen vermutlich an Schwelldosen aufgrund von körpereigenen Antioxidantien. Hierin ist auch eine mögliche Erklärung für die bescheidenen Ergebnisse von Sigurdsson zu sehen, wo nur mit geringen Leistungen unterhalb oder im Bereich dieser Schwellwerte gearbeitet wurde. Die beschriebene Wirkung für die Aknebehandlung in der WO 00/022491 liegt wahrscheinlich in einer oberflächlichen Bakterienabtötung durch den Blauanteil, wohingegen die Follikel im Wesentlichen nur durch die grünen bzw. roten Spektralbereiche von 520–550 nm bzw. 630–670 nm erreicht werden. Durch den Pulsbetrieb wird hingegen bei gleicher cw-Leistung temporär mit extrem höheren Leistungen gegenüber cw-Betrieb bestrahlt, so daß ein konstanter off-set aufgrund von Schwellwerten we-

niger stark ins Gewicht fällt. Somit erreicht effektiv mehr blaues Licht das Follikel und kann zur Bildung von Singulett-Sauerstoff beitragen.

[0010] Vorzugsweise liegen die effektiven Pulsängen zwischen 10 μ s und 10 ms besonders bevorzugt zwischen 100 μ s und 1 ms, wobei die Pulsein- und auszeiten unsymmetrisch sind. Unter effektiver Pulsänge wird dabei die Zeit verstanden, die zwischen Erreichen von 50% der maximalen Leistung bis zum Abfall auf 50% der maximalen Leistung liegt. Die im Verhältnis zur effektiven Pulsänge längeren Pulsauszeiten dienen dabei insbesondere der Nachdiffusion von Sauerstoff. Das Verhältnis liegt dabei vorzugsweise zwischen 10 und 100.

[0011] In einer weiteren bevorzugten Ausführungsform liegt die Frequenz, mit der die Bestrahlungsquelle gepulst wird, zwischen 0,1–1000 Hz, weiter bevorzugt zwischen 0,5–50 Hz und noch bevorzugter zwischen 1–10 Hz, wobei bei höheren Frequenzen niedrigere effektive Pulsängen und kleinere Pulsennergien verwendet werden.

[0012] Die Erzeugung längerer effektiver Pulsängen ist mit den bekannten Blitzlampen kaum oder garnicht realisierbar. Dies kann jedoch zur selektiven Anwärmung des Talgfollikelbereiches und des Haarschafts vorteilhaft sein, um eine Verflüssigung der obstruierenden Talgkonkremente und eine induzierte Störung der Talgproduktion zu erreichen. Ein weiterer positiver Effekt könnte eine Verminderung der Verhornung bzw. Abschilferung von Epithelzellen im Haarschaftsbereich sein. Derartige längere Pulsängen lassen sich jedoch in ihrer thermokinetischen Wirkung durch eine gezielte Ablaufsteuerung simulieren. Hierzu werden beispielsweise 100 Pulse mit einer effektiven Pulsänge von 100 μ s und einer Pulsauszeit von 900 μ s eingestrahlt, wobei anschließend für 10–1000 ms keine Pulse folgen. Anschließend werden dann wieder 100 Pulse eingestrahlt. Die hierbei verwendeten effektiven Pulsängen liegen zwischen 50–300 μ s.

[0013] In einer weiteren bevorzugten Ausführungsform ist die Bestrahlungsquelle als Xe-Blitzlampe ausgebildet. Diese handelsüblichen Xe-Blitzlampen sind sehr preiswert und emittieren ausreichend im gewünschten Spektralbereich zwischen 320–670 nm. Hierzu wird beispielsweise auf die US 4.167.669 oder die EP 0 565 331 verwiesen, wobei die dort beschriebenen Pulsennergien für die erfindungsgemäße Lehre jedoch zu groß sind. Xe-Blitzlampen sind je nach Belastung mehr oder weniger vom Spektrum mit einem Schwarzen Körper vergleichbar. Daher emittieren Xe-Blitzlampen typischerweise von 200–2000 nm. Aufgrund der Zelltoxizität der Wellenlängen zwischen 200–320 nm muß dieser Wellenlängenbereich durch entsprechende Filtermaßnahmen unterdrückt werden.

[0014] In einer weiteren bevorzugten Ausführungsform ist der Bestrahlungsquelle eine Einrichtung zur Unterdrückung der Spektralanteile von 320–400 nm und/oder zur Transformierung der UV-Anteile in den sichtbaren Bereich zugeordnet. Hiermit wird dem Umstand Rechnung getragen, daß die möglichen Nebenwirkungen von UVA-Anteilen auf die Zelle gänzlich vermieden werden, ohne daß sich dies in einer spürbaren Reduzierung der Wirksamkeit der Bestrahlungsanordnung niederschlägt. Werden die UVA-Anteile herausgefiltert, so können preiswerte handelsübliche UVA-Filter zur Anwendung kommen. Vorzugsweise werden jedoch die UV-Anteile mittels geeigneter Leuchtstoffe in den sichtbaren Spektralbereich transformiert. Dabei haben sich besonders aus Silikonelastomeren bestehende Folien mit anorganischen Leuchtstoffen bewährt. Aufgrund der hohen Absorption der Porphyrine um 420 nm werden bevorzugt blau emittierende Leuchtstoffe verwendet, die gegebenenfalls mit grünen im Bereich von 520–550 nm und/oder roten

im Bereich 630–670 nm kombiniert werden können.

[0015] Die Effizienz der Bestrahlungsanordnung kann weiter durch eine Erhöhung der Sauerstoffkonzentration erhöht werden. Neben den in der WO 00/02491 beschriebenen Maßnahmen kann dies auch sehr einfach durch eine inspiratorische Sauerstoffzufuhr über eine Sauerstoffmaske erreicht werden.

[0016] Aufgrund der eingestrahnten Pulsenergien kommt es zu einer spürbaren Erhöhung der Hauttemperatur, so daß vorzugsweise eine Kühleinrichtung für die zu bestrahlende Fläche vorgesehen ist. Diese kann im einfachsten Fall als Luftkühlung ausgebildet sein. Ebenso kann der Bestrahlungsquelle eine Kühleinrichtung zugeordnet sein, die beispielsweise als Luftkühlung oder eine andere thermische Ableitmaßnahme wie Kühlbleche ausgebildet ist.

[0017] Zur Erhöhung der abgestrahlten Leistung in Richtung der zu behandelnden Fläche wird die Strahlungsquelle vorzugsweise mit einem Reflektor ausgebildet. Ein bevorzugter Reflektortyp ist ein Paraboloid-Reflektor, wobei die Bestrahlungsquelle in einem Brennpunkt des Paraboloids angeordnet wird. Prinzipiell sind jedoch auch kugelschalenförmige oder ähnlich geformt Reflektoren verwendbar.

[0018] Die Bestrahlungsfläche für ein mobiles Gerät liegt vorzugsweise im Bereich von 1–200 cm², da bei flächiger Bestrahlung die Eindringtiefe im Vergleich zu punktförmigen Bestrahlungsquellen zunimmt, was hier zum Erreichen der tiefer liegenden Follikel vorteilhaft ist. Die mobile Ausführungsform erlaubt die sequentielle Bestrahlung verschiedener einzelner Akneherde, die in der Regel im Gesichtsbereich, im Halsbereich sowie im oberen Bereich des Rückens und des Brustkorbes auftreten.

[0019] Alternativ sind auch Ausführungsformen möglich, bei denen simultan größere Flächen bestrahlt werden. Eine mögliche Ausführungsform besteht darin, eine Vielzahl kleiner Blitzlampen, beispielsweise 30–60 kleine Xe-Blitzlampen in ein Gewebe einzunähen. Das besteht dabei beispielsweise aus PTFE oder einem PTFE-Derivat. Ein derartiges Gewebe kann durch direkte Metallbedampfung hochreflektiv beschichtet werden, wobei die erwünschte Luftdurchlässigkeit bei gleichzeitiger Abweisung von Wasser erhalten bleibt. Bei Verwendung einer Vielzahl von kleinen Strahlungsquellen in räumlicher Nähe zum Bestrahlungsbereich kann auf die Verwendung eines abbildenden Reflektors verzichtet werden. Mit Hilfe von weichen, strahlungstransparenten Abstandshaltern wie beispielsweise Silikonelastomeren ist eine Kühlung und Hinterlüftung der Filter, Leuchtstofffolien sowie der bestrahlten Hautflächen einfach über handelsübliche Lüfter wie beispielsweise CPU-Lüfter möglich.

[0020] Die Erfindung wird nachfolgend anhand eines bevorzugten Ausführungsbeispiels näher erläutert. Die Figur zeigen:

[0021] Fig. 1 einen Querschnitt durch eine Bestrahlungsanordnung,

[0022] Fig. 2 ein Spektrum der Bestrahlungsquelle mit und ohne Leuchtstoffolie und

[0023] Fig. 3 eine Haut-Querschnittsdarstellung mit Aknefollikel.

[0024] Die Bestrahlungsanordnung 1 umfaßt eine breitbandige Bestrahlungsquelle 2, die vorzugsweise als Xe-Blitzlampe ausgebildet ist. Die Bestrahlungsquelle 2 ist in einem Brennpunkt eines Paraboloid-Reflektors 3 angeordnet, der an der dem Brennpunkt abgewandten Seite offen ist. Die Austrittsfläche am offenen Ende des Paraboloidreflektors 3 wird durch eine vorzugsweise verstellbare Blende 4 definiert. Durch die verstellbare Blende 4 kann somit die Größe der zu bestrahlenden Fläche angepaßt werden. Die Bestrahlungsquelle 2 und der Paraboloid-Reflektor 3 sind in

einem Gehäuse 5 angeordnet. Das Gehäuse 5 ist vorzugsweise mit einem Handstück 6 ausgebildet, mittels dessen die Bestrahlungsanordnung 1 einfach auf eine zu behandelnde Fläche 7 aufsetzbar ist. Zwischen der Bestrahlungsquelle 2 und der zu behandelnden Fläche 7 ist eine Leuchtstofffolie 8 angeordnet, die mit Leuchtstoffpartikeln dotiert ist. Die Leuchtstofffolie 8 kann auch unmittelbar im Bereich der Bestrahlungsquelle 2 oder aber über die Blende 4 gespannt sein. Vorzugsweise ist die Leuchtstofffolie 8 derart angeordnet, daß diese leicht auswechselbar ist. Dies vereinfacht den notwendigen Austausch aufgrund von Alterungsprozessen, aber auch den flexiblen Einsatz von Leuchtstofffolien mit unterschiedlichen Leuchtstoffpartikeln. Des weiteren kann bei äußerer Anordnung der Leuchtstofffolie 8 diese leicht desinfiziert werden. Die elektrischen Anschlüsse und die Schaltung zur Erzeugung der variablen Pulsbreiten ist hier aus Übersichtsgründen nicht dargestellt.

[0025] In der Fig. 2 ist ein Spektrum einer verwendeten Xe-Blitzlampe mit und ohne Leuchtstofffolie dargestellt. Das Spektrum mit Leuchtstofffolie ist dabei gestrichelt dargestellt. Bei der Leuchtstofffolie handelt es sich um ein Silikonelastomer, das mit anorganischen Leuchtstoffen dotiert ist, die im blauen Spektralbereich von 400–450 nm bevorzugt emittieren. Die Leuchtstofffolie schneidet dabei den UV-Bereich zwischen 280–400 nm nahezu ab und transformiert diesen in den sichtbaren blauen Bereich von 400–450 nm.

[0026] Die Xe-Blitzlampe wird mit einer Frequenz zwischen 0,1–1000 Hz getaktet, wobei jedoch die effektiven Pulslängen nur zwischen 10 µs und 10 ms liegen. Die optischen Pulsenergien betragen dabei zwischen 1–3 J/cm².

[0027] Die Aknebehandlung erfolgt dabei über mehrere Tage bzw. Wochen, wobei die tägliche Behandlungsdauer zwischen 1 und 20 Minuten, vorzugsweise zwischen 2–5 Minuten liegt.

[0028] In der Fig. 3 ist ein Querschnitt durch die Haut im Bereich eines Haares 9 dargestellt. Das Haar 9 ist über einen verengten Ausführungsgang 10 mit einem mit Talg überfüllten und entzündeten Haarschaftsbereich 11 mit einer vergrößerten und entzündeten Talgdrüse 12 verbunden. Bei einem cw-Betrieb mit blauem Licht wird aufgrund der geringen Eindringtiefe (1/e) des blauen Lichtes sowie der Überwindung körpereigener Schwellwerte für blaues Licht bereits der überwiegende Anteil in der oberen Hautschicht oberhalb des Haarschaftsbereiches absorbiert, was schematisch durch den kurzen Pfeil 13 dargestellt ist. Beim Pulsbetrieb hingegen ist bei gleicher mittlerer Leistungsdichte die Leistungsdichte während des Pulses entsprechend dem Verhältnis der effektiven Pulslänge zur Frequenz wesentlich größer, so daß die konstante Abschwächung aufgrund der körpereigenen Schwellwerte geringer ins Gewicht fällt. Da somit die zur Verfügung stehende effektive Leistung größer ist, erreicht auch ein größerer absoluter Anteil des blauen Lichtes die tieferliegenden Haarschaftsbereiche 11 bzw. die Talgdrüse 12 und können dort zur lokalen Erzeugung von Singulett-Sauerstoff beitragen, was durch den längeren Pfeil 14 dargestellt ist.

Patentansprüche

1. Bestrahlungsanordnung zur Behandlung von Akne, umfassend mindestens eine Bestrahlungsquelle, **dadurch gekennzeichnet**, daß die Bestrahlungsquelle (2) ein breitbandiges optisches Spektrum im Bereich von mindestens 320 bis mindestens 670 nm emittiert, die Bestrahlungsquelle (2) im Pulsbetrieb betreibbar ist und die Pulsenergie kleiner 3 J/cm² und die gemittelte Leistungsdichte kleiner als 2 W/cm² ist.
2. Bestrahlungsanordnung nach Anspruch 1, dadurch

gekennzeichnet, daß die effektive Pulslänge zwischen 10 µs–10 ms liegt.

3. Bestrahlungsanordnung nach Anspruch 2, dadurch gekennzeichnet, daß die Bestrahlungsquelle (2) mit einer Frequenz von 0,1–1000 Hz getaktet ist.

4. Bestrahlungsanordnung nach einem der vorangehenden Ansprüche, dadurch gekennzeichnet, daß die Bestrahlungsquelle (2) als Xe-Blitzlampe ausgebildet ist.

5. Bestrahlungsanordnung nach einem der vorangehenden Ansprüche, dadurch gekennzeichnet, daß der Bestrahlungsquelle (2) eine Einrichtung zur Unterdrückung der Spektralanteile von 320–400 nm und/oder zur Transformierung der UV-Anteile in den sichtbaren Bereich zugeordnet ist.

6. Bestrahlungsanordnung nach Anspruch 5, dadurch gekennzeichnet, daß vor der Bestrahlungsquelle (2) eine Leuchtstofffolie (8) angeordnet ist.

7. Bestrahlungsanordnung nach Anspruch 6, dadurch gekennzeichnet, daß die Leuchtstofffolie (8) aus einem Silikonelastomer besteht und mit anorganischen Leuchtstoffpartikeln dotiert ist.

8. Bestrahlungsanordnung nach Anspruch 6 oder 7, dadurch gekennzeichnet, daß die Leuchtstofffolie (8) mit mindestens einem der nachfolgenden Leuchtstoffpartikeln, fluoreszierend in den Spektralbereichen 410–490 nm

[Sr₂P₂O₇:Eu, Sr₅(PO₄)₃Cl:Eu, BaMg₂Al₁₆O₂₇:Eu, CaWO₄:Pb; (Sr, Ca, Ba)₅(PO₄)₃Cl:Eu; Sr₂P₂O₇:Sn; (Ba, Ca)₅(PO₄)₃Cl:Eu]

und/oder 510–560 nm

[ZnSiO₄:Mn; MgAl₁₁O₁₉:Ce, Tb, Mn; YBO₃:Tb; LaPO₄:Ce, Tb]

und/oder 610–670 nm

[Y₂O₃:Eu; Y(P,V)O₄:Eu; CaSiO₃:Pb, Mn; (Sr, Mg)₃(PO₄)₂:Sn; 3.5MgO · 0.5MgF₂ · GeO₂:Mn] dotiert ist.

9. Bestrahlungsanordnung nach einem der vorangehenden Ansprüche, dadurch gekennzeichnet, daß der Bestrahlungsanordnung (1) Mittel zur topischen und/oder inspiratorischen Sauerstoffzufuhr zugeordnet sind.

10. Bestrahlungsanordnung nach einem der vorangehenden Ansprüche, dadurch gekennzeichnet, daß der Bestrahlungsanordnung (1) eine Einrichtung zur Kühlung einer zu bestrahlenden Fläche (7) zugeordnet ist.

11. Bestrahlungsanordnung nach einem der vorangehenden Ansprüche, dadurch gekennzeichnet, daß die Einrichtung als Luftkühlung ausgebildet ist.

12. Verfahren zur Behandlung von Akne mittels einer pulsablen, breitbandigen optischen Bestrahlungsquelle (2), die Pulse mit Pulsenergien kleiner 3 J/cm² in einem Wellenlängenintervall von 320 bis mindestens 670 nm erzeugt, wobei die mittlere Leistungsdichte kleiner als 2 W/cm² ist, umfassend folgende Verfahrensschritte:

a) Bestrahlen der Akne mit effektiven Pulslängen zwischen 10 µs–10 ms bei Pulsfrequenzen zwischen 0,1–1000 Hz mindestens 1–20 Minuten lang und

b) mindestens eine Wiederholung des Verfahrensschrittes a) nach mindestens 24 Stunden.

13. Verfahren zur Behandlung von Akne mittels einer pulsablen, breitbandigen optischen Bestrahlungsquelle, die Pulse mit Pulsenergien kleiner 3 J/cm² in einem Wellenlängenintervall von 400 bis mindestens 670 nm erzeugt, wobei die mittlere Leistungsdichte kleiner als 2 W/cm² ist, umfassend folgende Verfahrensschritte:

a) Bestrahlen der Akne mit effektiven Pulsfängen zwischen 10 μ s–10 ms bei Pulsfrequenzen zwischen 0,1–1000 Hz mindestens 1–20 Minuten lang und

b) mindestens eine Wiederholung des Verfahrensschrittes a) nach mindestens 24 Stunden. 5

14. Verfahren nach Anspruch 12 oder 13, dadurch gekennzeichnet, daß vor und/oder während der Bestrahlung die Sauerstoffkonzentration im Bereich der Akne durch topische und/oder inspirative Sauerstoffzufuhr 10 erhöht wird.

Hierzu 3 Seite(n) Zeichnungen

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Fig. 1

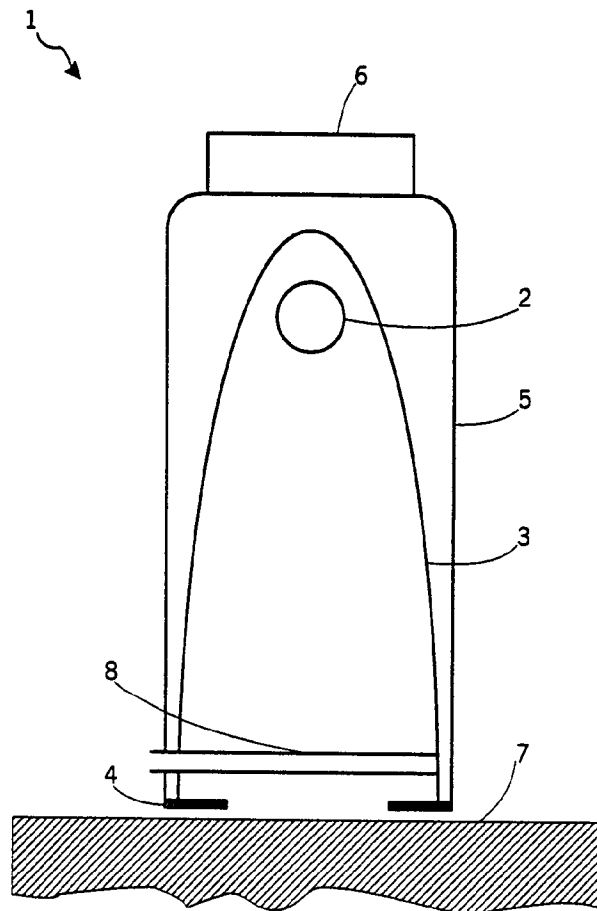


Fig. 2

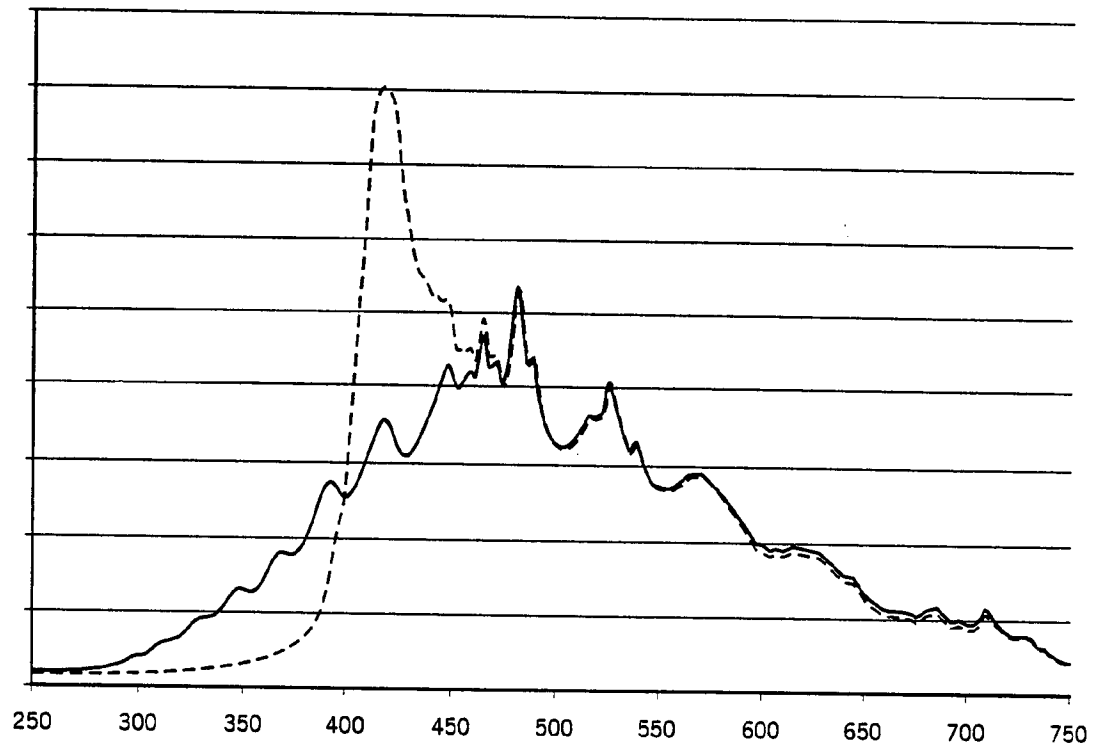
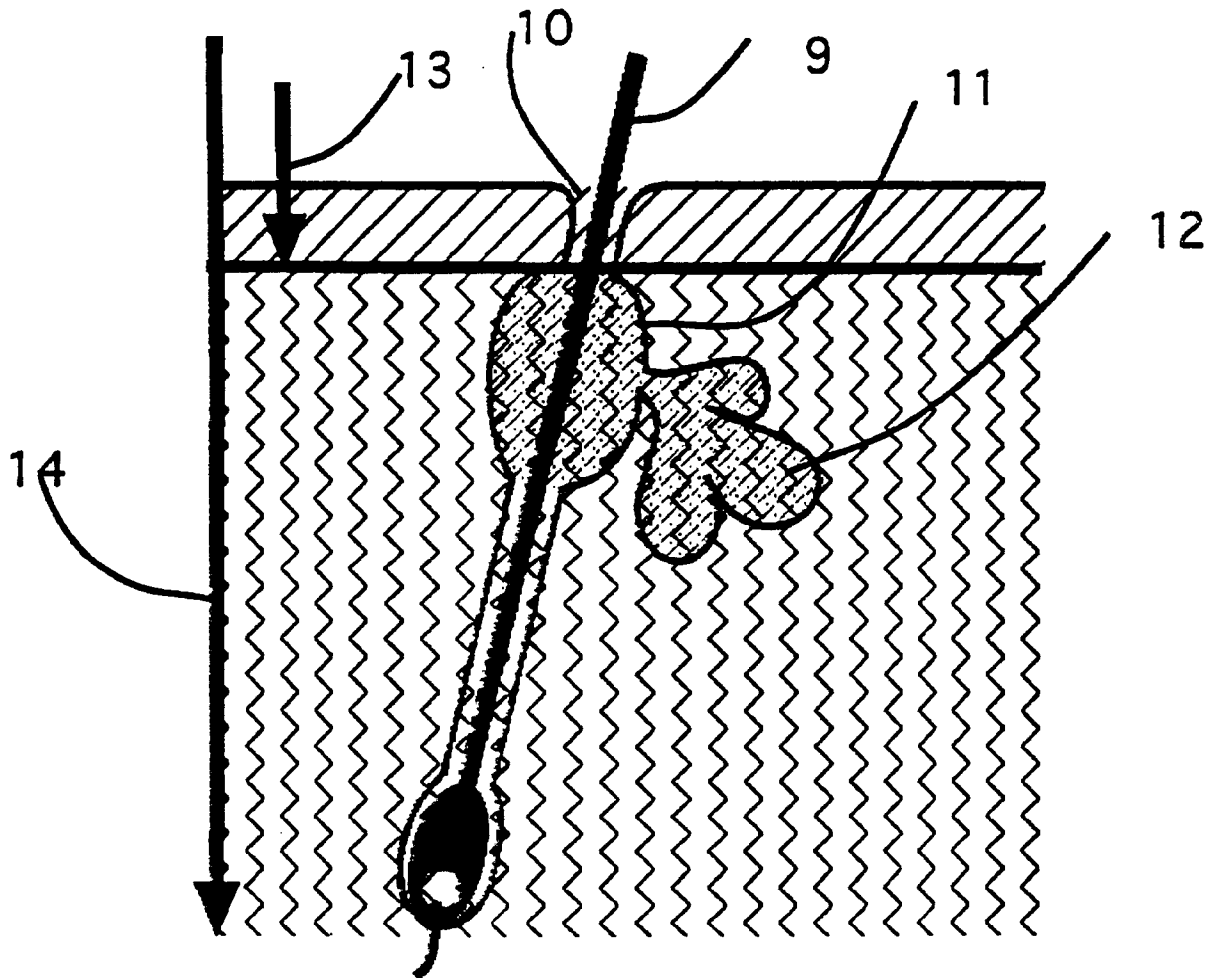
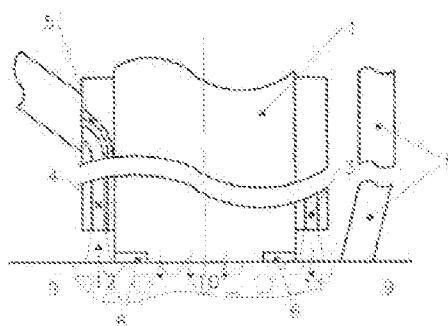


Fig. 3



Remission-controlled device with laser handpiece for sensor-controlled selective laser therapy of blood vessels and skin tissues has multiple-sensor system e.g. using near infrared or visible radiation**Publication number:** DE10120787 (A1)**Publication date:** 2003-01-09**Inventor(s):** SCHMIDT WOLF-DIETER [DE]; FASLER DIETER [DE];
SCHEIBE ARMIN [DE]; WOLLINA UWE [DE]**Applicant(s):** FOERDERUNG VON MEDIZIN BIO UND [DE]**Classification:****- international:** **A61B5/00; A61B18/20; A61B17/00; A61B18/00; A61B19/00;**
A61B5/00; A61B18/20; A61B17/00; A61B18/00; A61B19/00;
(IPC1-7): A61B18/22; A61B1/04; A61B5/01; A61B5/145;
A61B6/00; A61B8/00**- European:** A61B5/00P; A61B18/20H**Application number:** DE20011020787 20010425**Priority number(s):** DE20011020787 20010425**Cited documents:** DE2910760 (C2)
 DE19934038 (A1)
 US6165170 (A)
 US6156030 (A)
 US6015404 (A)[more >>](#)**Abstract of DE 10120787 (A1)**

A multiple sensor system in a concentric, polygonal or opposing form is arranged in immediate proximity to the laser beam outlet surface of the laser handpiece. The sensor system has at least two similar or different components, at least one of which uses remission spectroscopic principles, and preferably operates with visible and/or near infrared radiation.



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Prüfungsantrag gem. § 44 PatG ist gestellt

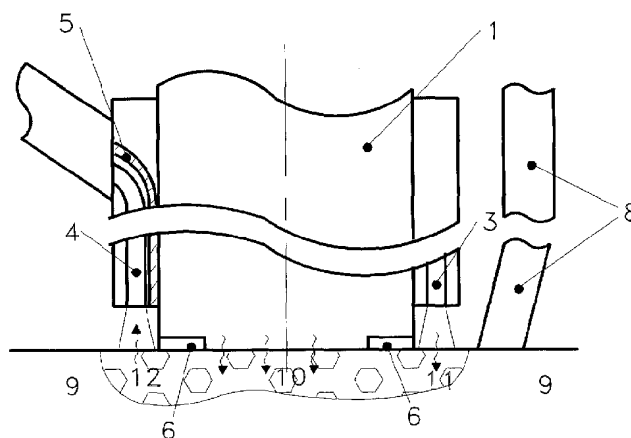
54 Anordnung zur remissionsgesteuerten, selektiven Lasertherapie von Blutgefäßen und Hautgewebe

57 Die Erfindung betrifft eine Anordnung für diagnostische Zwecke sowie zur Steuerung eines Lasertherapiegerätes.

An einem Laserhandstück ist in unmittelbarer Nähe der Laserstrahlaustrittsfläche in konzentrischer, polygonaler oder gegenüberliegender Form ein Mehrfach-Sensorsystem mit mindestens zwei gleichen oder unterschiedlichen Komponenten angeordnet.

Die Erfindung kann angewendet werden zur

- nichtinvasiven und nicht berührenden Erfassung von vaskulären Erkrankungen des menschlichen Hautgewebes,
- Steuerung von Lasertherapiegeräten für eine effektive und nebenwirkungsarme Behandlung von Varizen und anderen Erkrankungen des Hautgewebes,
- geometrischen Lagesteuerung eines Laserhandstückes über dem zu therapierenden Hautareal.



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[0001] Die Erfindung betrifft eine Anordnung zur remissionsgesteuerten, selektiven Lasertherapie zur Diagnostik von Blutgefäßen und Hautgeweben gemäß der Gattung des ersten Patentanspruches. Sie dient ferner zur Steuerung von Parametern eines Lasertherapiegerätes sowie zu dessen geometrischer Lagesteuerung über dem zu therapierenden Areal. Die Erfindung wird in Lasertherapiegeräten zur effektiven und nebenwirkungsarmen Behandlung von Varizen und anderen Erkrankungen des Hautgewebes eingesetzt.

[0002] Die therapeutische Wirkung von Laserstrahlung beruht bekanntlich auf Prozessen der Reflexion, Streuung und Absorption von Photonen, die sich innerhalb bestimmter, wellenlängenabhängiger Eindringtiefen im Gewebe ausbreiten können und in Absorptionszentren mit Melanin, Hämoglobin oder Wasser in Wärme umgewandelt werden. Sie erfolgt entweder lokal unspezifisch in großflächigen Gewebearealen oder spezifisch in Blutgefäßen. Ein wesentliches Absorbermaterial im Humangewebe ist Gewebewasser mit einem Absorptionsmaximum seiner Grundschwingung bei 2700 nm; Nebenmaxima existieren bei 1900 nm, 1460 nm und 980 nm. Oxihämoglobin hat drei Absorptionspeaks bei Wellenlängen von 420 nm, 548 nm und 575 nm. Das wichtigste Hautpigment Melanin weist eine, vom nahen infraroten (NIR) zum ultravioletten (UV) Bereich stark zunehmende Absorption auf.

[0003] Die Ausnutzung einer bestimmten Laser-Wellenlänge, um deren Strahlungsenergie in einem Targetchromophor wie Hämoglobin in Wärme umzusetzen und eine lokale Temperaturerhöhung von einigen zehn Grad bis zu einer Agglutination/Koagulation des Gefäßabschnittes zu erreichen, nennt man selektive Thermolyse. Bei der Lasertherapie vaskulärer Läsionen wurde bisher der gesamte Wellenlängenbereich von sichtbarer blau-grüner, gelber bis zu naher infraroter Strahlung angewandt. Dabei ist das Hämoglobin das wichtigste Targetchromophor. Das Melanin ruft jedoch besonders bei Laserwellenlängen im sichtbaren Spektralbereich als konkurrierender Absorber unerwünschte Nebeneffekte hervor. Die Pulsdauer der verwendeten Laser muss dabei auf die thermische Relaxationszeit im Zielgewebe abgestimmt sein, die von den jeweiligen Wärmeleitfähigkeiten und den spezifischen Wärmekapazitäten bestimmt werden. Bei zu kurzen Laserpulsen kommt es zu keiner ausreichenden Erwärmung des Gefäßabschnittes, bei zu langen zu einer Schädigung der umgrenzenden Gewebeareale. Zudem muss die Laserenergiedichte (Fluence) so gewählt werden, dass es weder zu explosionsartigen Verdampfungsprozessen im Gefäß kommt, die zu unerwünschten Rupturen mit anschließender Purpura-Bildung (Fluence zu hoch und Pulsbreite zu klein) führen, noch dass diese nur zu geringen Erwärmungen im Gefäßinneren führt, die ein erneutes Wachstum (Neoangiogenese) hervorrufen würde.

[0004] Die Lasertherapie wird am häufigsten bei Teleangiektasie (Erweiterung dünner Kapillargefäße), bei Feuermalen (angeborene kapillare Fehlbildung; PWS) und bei Hämangiomen (Blutgefäßgeschwulst) als Methode der Wahl eingesetzt, bei der Heilungsraten von 50% bis zu 90% erreicht werden. Eine erfolgreiche Behandlung lässt sich durch die optische Aufhellung der Haut erkennen, die neben einer herkömmlichen Farbfotografie besonders gut durch eine nicht-invasive, remissionsspektroskopische Untersuchung objektiviert werden kann. Die Effizienz einer Lasergefäßtherapie hängt von verschiedenen Faktoren ab: Die anatomische Lage, Durchmesser und Wanddicke der Gefäße sind im Zusammenhang mit dem verwendeten Lasertyp entscheidend für einen unmittelbaren Behandlungserfolg, der auch möglichst lang anhalten soll. Die Anzahl der erforderlichen Laserungen (ein- oder mehrmalig) spielt ebenfalls eine wichtige Rolle. Die physiologisch-morphologischen Hintergründe für eine erfolgreiche Gefäßtherapie ("Idealer Weg") sind noch Gegenstand wissenschaftlicher Grundlagenuntersuchungen. Histologische Untersuchungen haben z. B. ergeben, dass bei einer Besenreiser-Therapie mit grünem Laserlicht geeigneter Fluence und Pulsbreite eine Aggregation der Erythrozyten mit nachfolgender Thrombusbildung im Blutgefäß und Schrumpfung der Gefäßwände stattfindet. Die Funktionalität des Gefäßsystems umfaßt jedoch eine Vielzahl komplexer und interaktiver Variabler. In einem partiell zerstörten Blutgefäß werden eine Vielzahl von Wachstumsfaktoren und Inhibitoren wirksam, die eine Wiederherstellung des Gefäßes zum Ziel haben oder zu einer Angiogenese führen können. Dies kann eine mehrmalige, unmittelbar aufeinanderfolgende Laserbehandlung konterkarieren, da eine verstärkte Gefäßneubildung induziert wird. Das Laser-Therapieregime muss daher auf Form und Ausprägung der Varizen (Typisierung nach Weiss in der folgenden Tabelle) ausgerichtet werden, wobei zu berücksichtigen ist, dass der intravasale Anteil am Hauptchromophor Oxyhämoglobin geringer ausfällt als bei Teleangiektasien oder Gefäßmälnern (vaskuläre Naevi). Die charakteristischen Farbbilder und morphologischen Besonderheiten lassen sich für eine automatische optische Erkennung der Varizentypen verwenden:

		Ausprägung	Durchmesser	Farbe
Typ I	Teleangiektasien	intrakutan	0,1 bis 1 mm	rot
Typ Ia	"Matting"	intrakutan	<0,2 mm	flächig hellrot
Typ II	Venektasien		1 bis 2 mm	violett
Typ III	Retikuläre Venen	kurzstreckig varikös bei großen Venen	2 bis 4 mm	cyan-blau
Typ IV	Seitenastvarikosis		3 bis 8 mm	blau-grün
Typ V	Stammvarikosis	lange Strecken des Venenstamms	> 8 mm	

[0005] Für Lasertherapien eigneten sich bisher nur die Varizentypen I bis II [Bethge, S., Stadtler, R.: Der langgepulste frequenzverdoppelte Nd:YAG-Laser in der Behandlung von Besenreiser. Hautarzt 50, 181–185 (1999)]. In Abhängigkeit vom Durchmesser der varikösen Gefäße vom Typ I bis II geht man von zwei typischen klinischen Merkmalen (clinical

endpoint) aus, die zur Steuerung einer Laser-Applikation verwendet werden können:

Durchmesser	Optisches Merkmal der Blutgefäße nach erfolgreicher Lasertherapie
< 1 mm	sofortiges Verschwinden (ohne erneutes Refilling) oder blau-graue Verfärbung
1 bis 2 mm	blau-graue Verfärbung (Koagulation)

[0006] Die herkömmliche Sklerosierung variköser Gefäße mit Aethoxysklerol oder anderen zur lokalen Entzündung führenden Sklerosierungslösungen ist nach wie vor der Goldstandard in der vaskulären Therapie, obwohl sie als minimal-invasives Verfahren mit Belastungen für den Patienten verbunden ist. Der Therapieerfolg ist jedoch nicht immer zufriedenstellend; Komplikationen (Matting, Hyperpigmentierung, Nekrosen, Thrombosen und allergische Reaktionen) treten häufiger auf. Das Verfahren kann erst ab Gefäßdurchmessern größer als 1 mm sinnvoll eingesetzt werden. Ein Therapieerfolg der Sklerosierung wird im allgemeinen frühestens nach zwei Wochen sichtbar.

[0007] Die großen Erwartungen an die vaskuläre Lasertherapie sind jedoch in der Vergangenheit durch eine Reihe unbefriedigender Ergebnisse gedämpft worden. Die zunächst verwendeten kontinuierlich abstrahlenden blau-grünen Ar-Ionen-Laser (488 nm/514 nm) haben den Nachteil einer zu großen Bestrahlungsdauer, die auch durch einen mechanischen Shutter-Betrieb nicht unter 100 ms gehalten werden konnte. Gelbe Laserstrahlung von gepulsten Farbstoff-Lasern (580 nm) hatte sich zunächst bei der Lasertherapie von Gefäßerkrankungen gegenüber Argon-Lasern als überlegen erwiesen: Der Absorptionsdoppelpeak von Oxyhämoglobin wird mit gelbem Laserlicht besser getroffen und weniger Strahlung durch das epidermale Melanin unspezifisch in Wärme umgesetzt. [Alster, T. S. et al.: Dermatologic Laser Surgery. Dermatol Surg 22, 797–805 (1996)]. Allerdings ist die agglutinierende bzw. koagulierende Wirkung gelber Laserstrahlung im Blutgefäß nur auf eine Schichtdicke von unter 100 µm begrenzt: Die Absorption von Hämoglobin im Gefäß ist bei dieser Wellenlänge derartig stark, dass eine Abschirmung weiterer, darunter liegender durchbluteter Schichten bewirkt wird. Außerdem liegen die typischen Pulsbreiten von Farbstoff-Lasern unter 0,5 ms. Bei Gefäßdurchmessern größer als 0,1 mm wirkt daher gelbe Laserstrahlung wenig effektiv. Das Langzeitergebnis der Therapie verschlechtert sich, da die nur partiell geschädigten varikösen Gefäßbereiche sich regenerieren sowie Ausgangspunkt für ein erneutes Wachstum unerwünschter Gefäße werden können. Nebenwirkungen bei der Therapie mit blau/grünem und gelbem Laserlicht treten als epidermale Atrophie, Depigmentierung durch epidermale Melanozytenschädigung und Gefäßrupturen auf. Frequenzverdoppelnde Nd:YAG-Laser bei 532 nm lassen sich jedoch bei größeren Pulsbreiten (50 ms bei 20 J/cm²) betreiben und haben daher auch bei Gefäßen mit Durchmessern bis zu 1 mm eine erfolgreiche Therapie ermöglicht [Katz, B.: Laser therapy and sclerotherapy in treatment of large and small spider veins. Cosmetic Dermatology, September 1998]. Allerdings kann man grüne Laser nur bei hellhäutigen Patienten mit dem Hauttyp I und II anwenden; bei starker Bräunung (IV bis VI) sind sie aufgrund der starken Melaninabsorption und der zu erwartenden Nebenwirkungen nicht einsetzbar [Massey, R. A. et al.: Successful treatment of spider leg veins with a HELPFD Nd:YAG Laser. Dermatol Surg 25, 677–680 (1999)].

[0008] Besonders die langwelligere nah-infrarote Strahlung dringt tiefer in das Gewebe ein und kann damit auch größere Gefäße therapieren. Die Streuung der Photonen im Hautgewebe reduziert sich im NIR-Bereich gegenüber dem sichtbaren Spektralbereich um die Hälfte und ermöglicht daher deutlich größere Eindringtiefen. So wurde ein Nd:YAG-Laser (1064 nm) zur Therapie von Teleangiectasien und retikulären Venen mit Durchmessern bis 3 mm am Unterschenkel eingesetzt. Ein weiterer Vorteil dieser Wellenlänge ist neben der großen Eindringtiefe die vernachlässigbare Absorption von Melanin. Im Ergebnis der NIR-Laserung von Gefäßen tritt eine urtikarielle Schwellung auf. Der sichtbare Gefäßverschluss zeigt sich mit einer sofortigen Kontraktion und Abdunkelung. Es sollen keinerlei epidermale Schädigungen sogar beim Hauttyp IV sichtbar werden [Weiss, R. A.: Early clinical results with the multiple synchronized pulse 1064 nm laser for leg telangiectasias and reticular veins. Dermatologic Surgery 25; 399–402 (1999)].

[0009] Diodenlaser konnten aus technischen Gründen bisher kaum für die Lasertherapie eingesetzt werden, obwohl bereits erste Applikationsuntersuchungen an Teleangiectasien und Besenreisern bekannt geworden sind [Bass, L. S.: Photosclerosis of cutaneous vascular malformations with a pulsed 810 nm diode laser. Proceedings SPIE 2395, 559–565 (1995)]. Strahlung von Laserdioden bei 810 nm dringt ungefähr 2 mm in eine Blutschicht ein, so dass wesentlich dickere und tiefere Gefäße koaguliert werden können. Ein weiterer Vorteil der Laserwellenlänge bei 810 nm besteht darin, dass hier die spektrale Absorption von reduziertem und oxygeniertem Hämoglobin annähernd gleich ist (isobestischer Punkt). Ein Vaskularisations-Diodenlaser kann daher hier ohne Berücksichtigung des jeweiligen Sauerstoffsättigungsgrades der Gefäße eingesetzt werden. Bei venösen Unterschenkelvarizen liegt der Oxygenierungsgrad z. B. bei 70%, bei Feuermalen dagegen bei fast 100%. Auch die Wasserabsorption ist bei 810 nm gegenüber der im lokalen Maximum von 975 nm vernachlässigbar klein, so dass eine moderate Absorption von Strahlungsenergie durch das Hämoglobin im gesamten Gefäßdurchmesser erwartet werden kann. Da Melanin bei 810 nm jedoch auch noch eine signifikante Absorption aufweist, sollten durch eine geeignet eingestellte Kombination von Laserfluency und -Pulsdauer thermische Schädigungen der Epidermis vermeidbar sein. Erste therapeutische Anwendungen von Diodenlasern haben kaum Nebenwirkungen wie Narbenbildung und Hauttexturänderungen erbracht.

[0010] Ein Vergleich von inkohärenten Blitzlampenlichtquellen sowie FD-Nd:YAG-Lasern, Alexandrit-Lasern sowie von langwelligem Nd:YAG-Lasern (1064 nm) für die Gefäßtherapie steht noch aus. Es scheint sich jedoch zu bestätigen, dass inkohärentes Blitzlampenlicht keine breite Anwendung erfahren wird. [Goldberg, D. J.: Laser treatment of vascular lesions Clinics in Plastic Surgery 27; 173–180 (2000)]. [Goldberg, D. J.: A comparison of four frequency-doubled Nd:YAG (532 nm) laser systems for treatment of facial telangiectases. Dermatologic Surgery 25; 463–467 (1999)]; [Dover, A. U. et al.: New approaches to the treatment of vascular lesions. Lasers in Surgery and Medicine 26; 158–163 (2000)]; [Greve, B., Raulin, C.: Der Nd:YAG-Laser und seine Anwendungen in der Dermatologie. Hautarzt (51); 152–158 (2000)].

- [0011]** Ein wichtiger Aspekt bei lasertherapeutischen Verfahren ist die Vermeidung von zeitweiligen oder permanenten Nebenwirkungen. Dies betrifft besonders das Auftreten von punktförmigen Blutungen (Purpura), aber auch Wunden mit anschließender Narbenbildung, Hauttexturveränderungen und Pigmentverschiebungen bis hin zur Bildung von Plattenepithelkarzinomen. Die postoperative Bildung von Erythemen, Ödemen sowie von Hautbläschen wird in der Regel nicht als Nebenwirkung eingestuft. Allerdings sind sie als zusätzliche Komplikationen für den Patienten unerwünscht und sollten im Therapieverlauf möglichst vermieden werden. Die postinflammatorischen Hyperpigmentierungen werden durch den Blutabbau im koagulierten Gefäß (Hämosiderinablagerungen) sichtbar und sind im Verlauf einiger Monate reversibel. Die echten Nebenwirkungen zeichnen sich durch eine unspezifische thermische Laser-Schädigung von epidermalen und dermalen extravasalen Gewebsarealen aus. Besonders bei blau-grünen Lasern beobachtet man in der Epidermis und oberen Dermis eine unspezifische Koagulations-Nekrose, die häufig zu atrophischen oder hypertropischen Narben sowie zur Hypopigmentierung in der therapierten Haut führt [Haedersdahl, M. et al.: Cutaneous side effects from laser treatment of the skin. *Acta Derm Venereol* (Stockh) 78, 1–32 (1999)].
- [0012]** Eine optimale Lasertherapie vaskulärer Läsionen erfordert daher begleitende nichtinvasive Messverfahren, die auch eine on line-Steuerung des Lasergerätes ermöglichen sollen. Dies betrifft Sensoren für die Morphologie (laterale und vertikale Verteilung und Durchmesser) und Physiologie der Blutgefäße (relative Sauerstoffsättigung, Blutflussgeschwindigkeit). Hinzu kommt noch die Kenntnis der Melanin-Verteilung im Therapieareal. Für eine on line-Kontrolle einer Lasertherapie kommen möglichst nichtberührende Messverfahren in Betracht. Zur Gewinnung der primären Datenbasen für eine Therapieführung spielen neben Ultraschall- und fotoakustischen Methoden hierfür besonders die optische Kohärenztomografie, die Infrarot-Tomografie und die Remissionsspektroskopie eine wichtige Rolle.
- [0013]** Das Remissionsspektrum mit den wellenlängenabhängigen epidermalen und dermalen Streu- und Absorptionskoeffizienten, die vor und nach einer Lasertherapie ermittelt werden, enthält wesentliche vaskuläre (physiologische und anatomische) Informationen. Die präoperative Farbe des Hautareals, welches das zu behandelnde Gewebe mit der Gefäßveränderung aufweist, ist von großer Bedeutung (Bräunung!). Dies belegen erste Untersuchungsergebnisse von Remissionsspektren von lasertherapiertem Hautgewebe mit Feuernadeln deutlich [Sheehan-Dare, R. A. et al.: Copper vapour laser (578 nm) and FL pumped pulsed tunable dye laser (585 nm) treatment of port-wine-stains. *Br J Dermatol* 130, 478–482 (1994)]; [Troilius, A. et al.: Reflectance spectrophotometry in the objective assessment of dye laser-treated port-wine-stains. *Br J Dermatol* 132; 245–250 (1995)]; [von der Horst, C. M. A. M et al.: Effect of the timing of treatment of port-wine-stains with the FL pumped pulsed-dye laser *N Eng J Med* 338; 1028–1033 (1998)]. Über die Lösung des sogenannten inversen Problems sollen sich aus einem Remissionsspektrum auch die tiefenabhängige Verteilung der Blutgefäße und deren Durchmesser bestimmen lassen [von Gemert; M. J. C. et al.: Non-invasive determination of port wine stain anatomy and physiology for optimal laser treatment strategies. *Phys Med Biol* 42, 937–950 (1997)]. Voraussetzung dafür ist die Zusammenstellung einer Reihe typischer Remissionsspektren von Hautgeweben ohne und mit vaskulären Läsionen, von denen die anatomischen und optischen Eigenschaften z. B. über eine Histologie ermittelt wurden (Vorwärts-Problem). Vorläufige Ergebnisse weisen darauf hin, dass im Spektralbereich über 500 bis 600 nm Aussagen zum mittleren Durchmesser der Gefäße erhalten werden. Aufgrund der geringeren Eindringtiefe eignet sich der Spektralbereich unterhalb 500 nm zum Nachweis oberflächlicher Blutgefäße.
- [0014]** Spektroskopische Sensoren zur Steuerung von Medizinlasern bei der vaskulären Therapie sind bisher nicht bekannt geworden. Die prinzipielle Eignung von VIS-NIR-Spektalsensoren selbst für einen schnellen und berührungslosen Nachweis der epidermal/dermalen Mikrozirkulation konnte bereits aufgezeigt werden [Schmidt, W.-D., et al.: Non-contacting diffuse VIS-NIR spectroscopy of human skin for evaluation of skin type and time dependend microcirculation. *Proceedings of SPIE* 4160, 91–102 (2000)]; [DE 199 34 038 A1 (Schmidt/2001)]. Ein nicht-spektroskopisches, scannendes Farbsensorsystem (grün/rot) zur Gefäßtherapie wurde zur Erfassung der lateralen Verteilung der Blutgefäße im Gewebsareal ausgestattet, welches die on line-Steuerung der Laserparameter gestatten soll [Sebern, E. L., et al.: Design and characterization of laser-based instrument with spectroscopic feedback control for treatment of vascular lesion: the "smart scalpel". *J Biomed Optics* 5; 375–382 (2000)].
- [0015]** Bereits Anfang der neunziger Jahre wurde in den US-Patenten US 5 217 455 (Tan/1993) und US 5 312 395 (Tan/1994) der Einsatz von NIR-Lasern zur Therapie von vaskulären Läsionen beschrieben, wobei NIR-Laserwellenlängen von 600 bis 1100 nm, Fluencen von 1 bis 20 J/cm², Pulsdauern von 10 ns bis 300 ns sowie Strahldurchmesser der Laserapplikatoren von 1 bis 100 mm beansprucht worden sind. Laserpulsdauern < 300 ns waren allerdings nicht zur Behandlung vaskulärer Läsionen geeignet, allenfalls zur Zersprengung von subkutanen Farbpartikeln (Tattoos). Blitzlampengepumpte, grüne Nd:YAG-Laser (532 nm) bis zu 10 ms Dauer und 1 J/cm² sind für vaskuläre Anwendungen patentiert worden [US 5 558 667 (Yarborough/1996)]. NIR-Laser wurden zur selektiven Photothermolyse von Feuermalen (PWS) vorgeschlagen. Optimale NIR-Wellenlängen sollen bei 760 nm und 980 nm liegen [US 5 749 868 (Furumoto/1998)]. Auch gepulste Nd:YAG-Laser mit 1064 nm wurden zur kontrollierten thermischen Schädigung von Venen einbezogen [US 5 754 573 (Yarborough/1998); US 5 911 718 (Yarborough/1999)]. Raman-geshiftete Nd:YAG-Laser mit NIR-Wellenlängen bei 1,5 µm sollen Vorteile bei der Therapie tiefer Gefäße aufweisen, da sie im subepidermalen Gewebe effektiver das Wasser in den Blutgefäßen aufheizen und damit eine Koagulation bewirken können [US 5 897 549 (Tankovich/1999)]. Als allgemeine Methode einer NIR-Lasertherapie (800 bis 870 nm) wird das wiederholte Lasern, unterbrochen von einer jeweiligen Diagnose des Hautareals, beansprucht [US 5 464 436 (Smith/1995)]. Unmittelbar wiederholtes Lasern im gleichen Areal kann jedoch eine eher kontraproduktive Wirkung aufweisen. Die Verwendung von mehreren Wellenlängen in einem Laser-Therapiegerät entsprechend der jeweiligen vaskulären Läsion wurde in [DE 199 28 051 A1 (Katzner/2000)] beschrieben. Frühzeitig sind auch kühlende Anordnungen am Laserhandstück vorgesehen worden, mit denen durch Wasser, Kältespray oder thermoelektrische Mittel eine effektive Absenkung der Oberflächentemperatur der Epidermis (störende Melaninabsorption!) vorgenommen werden soll [US 5 282 797 (Chess/1994); US 5 486 172 (Chess/1996)].
- [0016]** Durch minimal-invasive Techniken (Lichtleiter, Endoskope etc.) kann Laserlicht auch in tieferliegende Gewebeteile und Organe als durch einfaches Einstrahlen über die Oberfläche eingebracht und für die kosmetische Chirurgie verwendet werden [US 5 505 727 (Keller/1996)]. Eine minimal-invasive Variante zur Therapie variköser Venen besteht

darin, mit einem CO₂-Laser oder einer Glasfaserspitze ein kleines Loch durch die Hautoberfläche bis zur entsprechenden Varize zu bohren und dann eine Koagulation des Blutgefäßes mit Laserstrahlung vorzunehmen [US 5 531 739 (Trelles/1996); US 5 552 813 (Trelles/1996); US 5 578 029 (Trelles/1996); US 5 733 277 (Pallarito/1998); US 5 807 385 (Keller/1998); US 5 984 915 (Loeb/1999)]. Abgesehen davon, dass hiermit kleinere Varizen kaum zu erfassen sind, erfordert dieses Verfahren eine Vielzahl einzelner Laserpunktionen pro Gefäßabschnitt, so dass lange Behandlungszeiten und eine hohe Infektionsgefahr die Folge wären. Aus der photodynamischen Laser-Therapie stammt der minimal-invasive Ansatz, vaskuläre wie neoplastische Läsionen lokal mit einem Photosensitizer anzureichern und anschließend mit Laserlicht zu bestrahlen, wodurch eine Photothrombose zu einem Gefäßverschluss führen soll [US 5 558 667 (Yarborough/1996)]. Ein Vorteil dieser Therapie sollen niedrigere Fluencen im Vergleich zur nichtinvasiven Lasertherapie sein; allerdings kann auch hier ein erneutes Wachstum der Gefäße nicht ausgeschlossen werden.

[0017] Auch Diodenlasersysteme sind bereits für den Einsatz bei vaskulären Läsionen vorgeschlagen worden, wobei erfindungsgemäß mit Laserstrahlung mit Wellenlängen von 700 bis 1100 nm, Fluencen von 5 bis 100 J/cm² und Pulsdauern bis zu 100 ms gearbeitet werden soll [US 5 658 323 (Miller/1997); US 5 707 403 (Grove/1998); US 5 879 376 (Miller/1999); US 6 027 495 (Miller/2000); US 6 096 029 (O'Donnel/2000); US 6 149 644 (Xie/2000)]. Gepulste NIR-Laser werden zur Behandlung von Hautkrankheiten wie Psoriasis vorgeschlagen. Durch die große Eindringtiefe der NIR-Strahlung sollen abnormale Blutgefäße unter psoriatischem Plaque selektiv bis in eine Tiefe von 1 mm beseitigt werden [US 5 527 350 (Grove/1996)]. Infrarot-Laser lassen sich zur Gefäßtherapie des Augenhintergrundes einsetzen, wobei aufgrund des transparenten Augeninneren eine relativ einfache Strahlungseinkopplung sowie Einschätzung des Therapieerfolges über eine Video-Kamera erfolgen kann, wobei das Videobild eine einfache Steuerung des Therapie-Lasers ermöglichen soll [US 5 400 791 (Schlier/1995)]. Ein ähnliches Konzept wird zum automatisierten Lasertracking in der Gefäßtherapie wie bei Besenreisern vorgeschlagen, wobei eine CCD-Kamera die variköse Gefäßstruktur aufnimmt und mittels der Bildinformationen die manuelle/programmgesteuerte Positionierung des Laserstrahls im Therapieareal ermöglicht werden soll [US 5 653 706 (Zavitslan/1997); US 5 860 967 (Zavitslan/1999)]. Im US-Patent [US 6 015 404 (Altshuler/2000)] wird ein dermatologisches Lasersystem mit einer sogenannten Feedback-Kontrolle durch einen akustischen Temperatursensor, der sich im Laserhandstück in Kontakt mit der Haut befindet, beschrieben. Mit einem ersten Laserpuls geringer Fluence soll zunächst eine Vordiagnostik des zu therapierenden Hautareals durchgeführt werden; die größere Fluence des zweiten Laserpulses wird entsprechend eingestellt und sorgt für die therapeutische Wirkung. Damit soll eine on line-Kontrolle des Lasergerätes möglich sein. Eine Methode zur Kontrolle der Hauttemperatur bei der Photokoagulation von Gefäßen mit einem Nd:YAG-Laser besteht in der Strahlungstemperaturmessung der Haut, womit die Kältespray-Applikation gesteuert werden soll [US 5 979 454 (Anvari/1999)]. In einem weiteren Patent wurde zur Hauttemperaturüberwachung ein Thermoelement im Laserhandstück (in Kontakt mit der Haut) verwendet [US 6 096 029 (O'Donnel/2000)]. Nachteilig an diesem Verfahren sind sowohl die hohen Zeitkonstanten als auch die Wärmeübergangswiderstände, die keine genaue Temperaturentwertung im Gefäßbereich erlauben.

[0018] Spektroskopische Methoden zur Charakterisierung von biologischen Geweben beruhen auf dem Nachweis der Remission im ultravioletten, sichtbaren und naheninfraroten Spektralbereich sowie der Fluoreszenz. Der Nachweis von Gewebeschädigungen bei einer Lasertherapie mittels optisch-spektroskopischer Methoden wurde allgemein in [US 6 015 404 (Altshuler/2000)] beansprucht. Ein ähnlicher spektralsensorischer Ansatz wird auch aus der Patentschrift [DE 199 28 051 A1 (Katzner/2000)] deutlich, wo zunächst spektral aufgelöste Intensitätswerte der Hautareale vor einer Lasertherapie ausgewertet werden. Aus dem Vergleich mit spektralen Datensätzen von vaskulären Läsionen wird dann die optimale Wellenlänge von Gefäß-Lasern ausgewählt, die zur Therapie eingesetzt werden soll. Ein spektroskopisches Gerät und Verfahren zur berührungslosen, abstandskontrollierten Remissionssensorik für Hautgewebe wurde in [DE 199 34 038 A1 (Schmidt/2001)] dargestellt. Zur Steuerung eines UV-Ablationslasers soll die charakteristische Remission der Haut, besonders der Pigmentierung und der Blutgefäße, mit einem weiteren Spektralsensor verwendet werden [38 US 6 165 170 (Wynne/2000)]. Die hier beschriebenen optischen Merkmale (Hautfarbe; Auftreten von Punktblutungen) dienen allerdings der Steuerung der Ablationstiefe eines UV-Lasers und haben keinen Bezug zur nichtinvasiven vaskulären Therapie. Weiterhin wird zur Steuerung eines Ablationslasers bei Brandverletzungen eine spektroskopische Methode vorgeschlagen, mit der über Fluoreszenz- und Remissionsmessungen zwischen unverletztem und nekrotischem Gewebe unterschieden werden soll [US 5 701 902 (Vari/1998)]. In einem Endoskop wird in Ergänzung zur Biopsie eine spektroskopische Kontrolle vorgesehen, die auch bei vaskulären Erkrankungen zum Einsatz kommen soll [US 5 843 000 (Nishioka/1998)]. Die remissionsspektroskopische Bestimmung von Blutbestandteilen wie Glucose wird in [US 5 935 062 (Messerschmidt/1999)] beansprucht. Zur Auswertung von Remissionsspektren wird als allgemeine Methode die Verwendung von spektralen Datensets und Look-Up-Tafeln angegeben [US 6 115 673 (Malin/2000)].

[0019] Der Erfindung liegt die Aufgabe zugrunde, mit einer Anordnung eine sensorgesteuerte, selektive Lasertherapie von varikösen Blutgeweben und pathologischen Hautgeweben zu ermöglichen. Gemäß der Erfindung wird diese Aufgabe durch das Kennzeichen des ersten Patentanspruchs gelöst. Es wird also ein Laserhandstück mit einem Mehrfach-Sensorsystem zur Diagnostik von Blutgefäßen und Hautgeweben angeordnet, dessen Ausgangssignale zur Steuerung von Parametern eines Lasertherapiegerätes wie der Fluence, der Pulsbreite und der Pulsfrequenz verwendet werden. Dadurch wird es möglich, eine auf das Therapieareal, angepasste Laserbestrahlung vorzunehmen und thermische Nebenwirkungen im Gewebe zu vermeiden. Außerdem werden die Sensorsignale zur geometrischen Lagesteuerung des Laserhandstücks über dem zu therapierenden Hautareal abgewandt. Dadurch gelingt eine exakte Positionierung des Laserhandstücks über großen und flächigen Varizen, und man vermeidet eine schädliche Bestrahlung normaler Hautgewebe oder bereits bestrahlter Hautareale. Das Kernstück des Mehrfach-Sensorsystems besteht dabei aus einer oder mehreren spektroskopischen Messeinrichtungen, vorzugsweise im sichtbaren und nahen-infraroten Wellenlängenbereich. Darüber hinaus sind Temperatur-, Druck-, Ultraschall- und Bildaufnahmesensoren in das Laserhandstück integriert. Außerdem ist die erfindungsgemäße Sensorlösung als Ergänzung von Geräten zur dauerhaften Haarentfernung (Epilationslaser) anwendbar.

[0020] Die erfindungsgemäße Lösung zur sensorgesteuerten selektiven Lasertherapie von Blutgefäßen und Hautgeweben ist für die Steuerung und Regelung von medizinischen Lasergeräten geeignet. Herkömmliche Sklerosierungsverfahren

ren mittels minimal-invasiver Injektionen in variköse Gefäße sind im allgemeinen schwierig bei kleinen Durchmessern zu handhaben und erlauben nicht immer befriedigende Therapieergebnisse. Als Nebenwirkungen können Pigmentverschiebungen, Wunden und Narben auftreten. Häufig tritt ein teleangiektatisches Matting als kapillare Aussprossung kleinster Gefäße auf. Auch bei einer herkömmlichen Sklerotherapie großer Varizen werden nicht alle weitverzweigten

ektatischen Gefäße erreicht. Daher hat sich die Lasertherapie vaskulärer Erkrankungen zur Methode der Wahl entwickelt.

[0021] Nachteile bekannter Lasergeräte und Laserhandstücke bestehen besonders in nicht an die Gefäßerkrankung angepassten Laserparametern, wie der Wellenlänge, der Fluence und der Laserpulsdauer. Sie zeigen sich als thermische Schädigungen der Epidermis und Dermis sowie in einem wenig effizienten Therapieverlauf. Laserwellenlängen im sichtbaren Bereich haben eine zu geringe Eindringtiefe, um auch tieferliegende Varizen zu erreichen. Auch die unmittelbare Ausrichtung der Wellenlänge auf das Hämoglobin als wichtigstem Targetchromophor bei gelbem Laserlicht hat sich für die Gefäßtherapie als ungünstig herausgestellt, da durch die starke Strahlungsabsorption im Blut meist nur die oberen Abschnitte des Gefäßdurchmessers koaguliert werden und eine ungewollte Regeneration des Gefäßes wahrscheinlich ist. Bei noch kurzwelligerem blau-grünem Laserlicht ist die Strahlungsabsorption in der Epidermis so stark, dass hier thermische Nebenwirkungen unvermeidlich sind, die mitunter auch durch eine aufwändige Kühleinrichtung nicht vermieden werden können. Daher wird die Notwendigkeit einer spektralen Einschätzung des Bräunungszustandes der Haut deutlich. Andererseits kann die Notwendigkeit einer Kühleinrichtung sich bereits als nachteilig für die vaskuläre Therapie erweisen, da durch die Kühlung der Haut nicht nur die Epidermis betroffen ist, sondern auch die Dermis mit den zu therapierenden Gefäßen auf Temperaturen abgekühlt wird, die unter Umständen auch bei hohen Laserfluencen keine agglutinierenden Wirkungen mehr erlauben. Eine Nichtbeachtung der thermischen Relaxationszeiten von Hautgewebe mit Gefäßen führt bei zu geringer Laserpulsbreite und hoher Fluence zum Platzen der Blutgefäße mit Purpura-Folge sowie bei zu langer Pulsbreite zur thermischen Schädigung umliegender Hautgewebe. Bei gekühlten Laserhandstücken, die im Kontakt mit der Hautoberfläche angewendet werden, kann ein versehentlicher Kontaktverlust ebenfalls zu thermischen Nebenwirkungen führen.

[0022] Erfindungsgemäß wird daher eine Messanordnung von einem Therapielaser, bevorzugt von einem Diodenlaser, mit einem Handstück zur sensor- und remissionsgesteuerten, selektiven Lasertherapie von Blutgefäßen und Hautgeweben vorgeschlagen. Diodenlaser sind kompakte, portable und zuverlässige Therapiegeräte. Die Strahlung von Diodenlasern mit einer NIR-Wellenlänge bei 810 nm besitzt eine große optische Eindringtiefe in Hautgewebe bis in den Bereich von 5 bis 10 mm. Durch die moderate Absorption von Hämoglobin kann das Targetchromophor gut zur tiefgründigen Agglutination und Koagulation in Blutgefäßen ausgenutzt werden, während die Absorption des störenden Pigments Melanin bereits stark reduziert ist. Die Wasserabsorption, die eine unspezifische Wirkung in allen Gewebeteilen ausweist, bleibt bei dieser Wellenlänge hinreichend klein. Durch eine Mehrfach-Sensoranordnung lassen sich gefäß- und hautgewebespezifische Parameter erfassen, die für eine Steuerung der Laserparameter sowie für eine Lagesteuerung des Laserhandstückes über dem zu therapierenden Areal verwendet werden. Die Verwendung der sensorgesteuerten Diodenlaseranordnung erlaubt bei der Gefäßtherapie eine deutliche Erweiterung behandelbarer Varizen- und Hauttypklassen. Das Schmerzempfinden bei und nach der Lasertherapie wird ebenfalls deutlich reduziert. Durch die geringere Melanin-Absorption wird eine wesentlich geringere Nachkühlzeit der therapierten Hautflächen im Vergleich zu den heute üblichen 20 Minuten (bei grünem Laserlicht) ermöglicht. Auch dadurch erreicht man eine effektivere Behandlung.

[0023] Erfindungsgemäß wird die Aufgabe durch ein Laserhandstück mit einem Mehrfach-Sensorsystem realisiert, das als wesentliche Komponente einen oder mehrere Spektralsensoren enthält. Die Spektralsensoren sind als miniaturisierte Spektrometermodule mit Silizium-Zeilen für den sichtbaren Wellenlängenbereich von 400 bis 1000 nm sowie mit InGaAs-Zeilen für den nahen infraroten Bereich von 1000 bis 2500 nm ausgeführt. Sie werden über eine spezielle Zeilenelektronik angesteuert und ausgelesen. Ein PC-gestütztes Auswertesystem bereitet die Remissionssignale nach bekannten Verfahren auf und setzt sie erfindungsgemäß in Steuersignale für das Lasergerät um. Derartige remissionsspezifische Auswertealgorithmen betreffen die Remissionsänderungen von einem nichttherapierten Gefäßabschnitt zu einem koagulierten (Farbumschlag von blau-rot zu grau-weißlich) oder Remissionsunterschiede zwischen krankhaftem und gesundem Hautgewebe zur exakten Positionierung des Laserhandstücks. Das Laserhandstück soll dabei sowohl in einer berührenden wie auch nichtberührenden Betriebsweise über der Haut eingesetzt werden können. Im berührenden Betriebsfall kann mit einem integrierten Kontaktkühler die restliche störende Wärme aus der Epidermis abgeführt werden. Da durch den Kontaktdruck Blut aus den Gefäßen gepresst wird, welches zur effizienten Lasertherapie benötigt wird, erfolgt die remissionsspektroskopische Sensorik erfindungsgemäß vorzugsweise berührungslos, indem die entsprechenden Lichtleiter kreisförmig um die Laserlichteinrichtung sind, jedoch in einem Abstand von einigen Millimetern oberhalb der Hautoberfläche angebracht sind. Dadurch wird gewährleistet, dass die varikösen Gefäße mit einer weitgehend ungestörten Mikrozirkulation detektiert werden können. Die Einhaltung eines konstanten Abstandes über der Haut wird in bekannter Weise im einfachsten Fall über einen mechanischen Abstandshalter oder mittels einem weiteren NIR-Remissionssensor einem oder Laser-Triangulationssensor vorgenommen.

[0024] Erfindungsgemäß wird die spektrale Remission über eine Lichtleiteranordnung, die an die Form der Laserlichteinrichtung angepasst ist, aufgenommen. Dabei sind z. B. kreisförmig abwechselnd die Lichtleiter zur Bestrahlung mit Halogenlampenlicht sowie zur Detektion des remittierten Lichts angeordnet. Dadurch wird bei einer Bewegung des Laserhandstücks über ausgedehnten Gefäßen die spektrale Remission sowohl vor als auch nach einer Laserbestrahlung nachweisbar. Durch Auswertung der Remission im noch nicht therapierten Areal können Gefäß- und Hauteigenschaften rechnerisch bestimmt und als Steuerparameter für das Lasergerät verwendet werden. Diese Steuersignale werden entweder über ein separates Display am Lasergerät angezeigt und dienen dem Operator als unterstützender Hinweis oder bewirken im Lasergerät selbst eine Veränderung von Fluence, Pulsbreite oder Frequenz in Richtung auf eine effizientere und nebenwirkungsärmere Therapie.

[0025] Gemäß der Erfindung wird auch der Bräunungszustand der Haut remissionsspektroskopisch erfasst, als Melanin-Index quantitativ ausgedrückt und zur Steuerung der Laseranlage verwendet werden. Eine präoperative Bräunung kann ein Ausschlussgrund für eine Lasertherapie sein. Zudem steigen bei starker Bräunung durch eine nichtabge-

stimmte Laserwellenlänge und Fluence die Schmerzempfindungen und die Nebenwirkungsgefahr an. Die VIS-NIR-Spektralsensorik gestattet darüber hinaus die Bestimmung physiologischer und morphologischer Gefäßparameter wie des relativen Wassergehalts, der relativen Sauerstoffsättigung sowie der Gefäßstruktur. Für die rechentechnische Auswertung der Spektren werden Haut-Gefäß-Modelle verwendet, die als inverses Problem für verschiedene Krankheitsbilder der Haut vorhanden sind oder geschaffen werden können.

[0026] Ein weiterer Grundgedanke der Erfindung besteht in der Anordnung zusätzlicher Sensoren im Laserhandstück wie u. a. für Temperatur (Silizium-Widerstandssensor, Thermistor), Feuchte (kapazitiv, FET) und Kontaktdruck (Silizium-Sensor), mit denen beim berührenden Laserhandstück entsprechende Messwerte überwacht und mit Referenzwerten verglichen werden, die für eine effiziente und nebenwirkungsarme Therapie erforderlich sind. Diese sind entweder wie die Hauttemperatur bekannt oder werden wie der Kontaktdruck und Feuchte in Vorversuchen ermittelt und in der elektronischen Auswerteeinrichtung zur Referenzbildung abgespeichert. Bei einem nichtberührenden Laserhandstück lässt sich die Gewebetemperatur über einen Strahlungstemperaturmesser (Thermopile, pyroelektrischer Sensor) bestimmen. Dieses Messverfahren hat außerdem den Vorteil, dass nicht nur die Oberflächentemperatur sondern auch thermische Effekte in tieferen Schichten einbezogen werden.

[0027] Die Erfindung ist weiterhin durch eine Messanordnung gekennzeichnet, die aus einem gepulsten Hochleistungs-Diodenlaser mit rotem Zielstrahllaser, vorzugsweise mit einer Wellenlänge von 810 nm, einer Frequenz bis 4 Hz, einer variablen Energiedichte bis 64 J/cm², einer variablen Pulsbreite bis 60 ms und einer Laserspotgröße bis zu 12 mm besteht. Die Lichtleiteinrichtung zur Laserstrahlzuführung ist vorzugsweise aus einem optischen Hohlraum-Leitersystem oder aus einem Bündel von LOH-Quarzglasfasern aufgebaut. Die Beleuchtungsquelle für die Remissionsuntersuchungen besteht aus einer 100 W-Wolfram-Halogenlampe, die aus einem stabilisierten Netzteil gespeist wird. LOH-Quarzlichtleiter sorgen für Lichtleitung zum Gewebe und leiten das remittierte Licht zu den Spektrometermodulen. Dabei sind für den VIS- und NIR-Bereich unterschiedliche Spektrometermodule vorgesehen, die von den jeweils n Detektor-Lichtleitern entweder im Multiplex angesteuert werden oder im Parallelbetrieb in n-facher Anzahl vorhanden sind. Die miniaturisierten Spektrometermodule haben vorzugsweise eine spektrale Auflösung von 5 nm im VIS-Bereich und von 10 nm im NIR-Bereich. Weiterhin ist über ein IR-Lichtleiterbündel (z. B. aus As-Te-Glas) ein integrierter Temperaturstrahlungsdetektor realisierbar. Schließlich sorgt eine Auswerte-Steuerungs-Elektronik (Build-In-Computer aus PC-Card- oder Mikrorechner-Baugruppen) für eine Steuerung des Lasergerätes, die Kommunikation zwischen den Baugruppen und die Auswertung der Signale des Mehrfachsensorsystems. Mehrere Stromversorgungsbaugruppen stellen die nötigen Betriebsspannungen bereit.

[0028] Ein weiterer Grundgedanke der Erfindung ist, dass die Laserstrahlungsquelle und die Halogenlampenstrahlung über einen optischen Umschalter (Shutter) abwechselnd in die zentrale Lichtleiteinrichtung des Laserhandstücks eingekoppelt werden. Dadurch lassen sich vorteilhaft die remissionsspektroskopisch beobachteten Hautareale mit den zu therapierenden kombinieren. Weiterhin wird auf diese Art auch die Ankopplung einer CCD-Videokamera ermöglicht, wodurch in der Bestrahlungsphase mit der Halogenlampe ein Bild vom zu therapierenden Gewebe aufgezeichnet und nach einer Computerbildanalyse zur Lasersteuerung verwendet werden kann.

[0029] Gemäß der Erfindung werden die Signale der Mehrfach-Remissionssensoranordnung von größeren varikösen Gefäßen zur Anzeige der Verfahrrichtung des Laserhandstücks mittel einer mehrteiligen LED-Indikatoreinrichtung verwendet. Dies wird durch eine Ausnutzung unterschiedlicher spektraler Merkmale von varikösen Gefäßen, gelaserten Gefäßen sowie gesundem Gewebe erreicht. Ein eingeschaltetes Indikatorelement (z. B. eine LED) soll dabei dem Operator die Verfahrrichtung des Laserhandstücks anzeigen.

[0030] Die Erfindung soll nachstehend anhand eines Ausführungsbeispiels erläutert werden.

[0031] Die Zeichnungen zeigen:

[0032] Fig. 1 Seitenansicht des Laserhandstückes mit Mehrfach-Sensorsystem

[0033] Fig. 2 Draufsicht des Laserhandstückes entsprechend Fig. 1

[0034] Fig. 3 Unteransicht des Laserhandstückes entsprechend Fig. 1

[0035] Fig. 4 Darstellung der Messanordnung

[0036] Das dieser Erfindung zugrunde liegende sensorgesteuerte Laserhandstück erfüllt wesentliche medizinisch-technische Anforderungen an eine selektive Lasertherapie von Blutgefäßen sowie Haut- und Wundgeweben. Dies betrifft die Durchführung effizienter und nebenwirkungsarmer, lasertherapeutischer Maßnahmen an Patienten, die dabei nur einer geringen Schmerzbelastung ausgesetzt sein sollen.

[0037] In den Fig. 1, 2 und 3 sind schematische Darstellungen des Laserhandstücks zur remissionsgesteuerten, selektiven Lasertherapie von Blutgefäßen und Hautgeweben mit Mehrfach-Sensorsystem in verschiedenen Ansichten dargestellt, die als sensorisches Kernstück eine remissionsspektroskopische Anordnung von Lichtleitern enthalten. Das Laserhandstück besteht aus einem portablen Gehäuse 1 mit einer seitlich angeordneten Kontaktkühleinrichtung 8. Die Laserstrahlungsübertragungseinrichtung 2 besteht im wesentlichen aus einem zylindrischen Quarzkörper mit einem typischen Durchmesser von 10 mm und liegt auf dem zu therapierenden Hautareal auf. Zur Verbesserung der optischen und thermischen Ankopplung zwischen Laserhandstück und Hautgewebe (Laserphotonen 10) kann das Hautgewebe 9 mit einer dünnen Schicht Ultraschallgel überzogen sein. An der Lichtaustrittsfläche des Laserhandstückes befinden sich Halbleiter-Drucksensoren 6, die zur Einhaltung eines Kontaktdruckes des Laserhandstückes zwischen einem experimentell ermittelten oberen und unteren Grenzwert verwendet werden. Weiterhin ist an dieser Lichtaustrittsfläche ein Kontakttemperatursensor 7 zur Überwachung der oberflächlichen Hautgewebetemperatur angeordnet. Mit ihm soll die Überschreitung einer kritischen Hauttemperatur signalisiert werden, die durch eine zu große Laserstrahlungsleistung zu dauerhaften Epidermisschädigungen führt. Ringförmig um die Laserübertragungseinrichtung 2 ist eine Halogenlampen-Strahlungsübertragungseinrichtung 3 angeordnet, die aus einer größeren Anzahl von LOH-Quarzlichtleiterfasern besteht. Die Lichtleiter weisen eine hohe Transmission im Wellenlängenbereich von 400 bis 2500 nm auf. Dazwischen ist abwechselnd die ringförmige Remissions-Strahlungsübertragungseinrichtung 4 ebenfalls aus LOH-Quarzlichtleiterfasern angeordnet, so dass mit Glasfasern im Millimeter-Abstand das Hautgewebe 9 mit Halogenlampenlicht bestrahlt (emittierte Photonen 11) und das remittierte Licht aufgenommen werden kann (remittierte Photonen 12). Die Anzahl der jeweils an-

gebrachten Sende- und Detektor-Lichtleiter soll möglichst groß sein und richtet sich nach den technologischen Fertigungsmöglichkeiten. Die Lichtleiteranordnungen **3** und **4** sind seitlich konzentrisch an der Laserübertragungseinrichtung **2** derartig angebracht, dass durch einen Abstand von einigen Millimetern über dem Hautgewebe eine berührungslose Messung der Remission ermöglicht wird. Dadurch sollen Verfälschungen durch den Kontaktdruck des Laserhandstückes vermieden werden. Durch die konzentrische Anordnung der Remissionslichtleiter **4** kann weiterhin bei beliebiger Verfahr-
 richtung des Laserhandstückes auf dem zu therapierenden Hautareal gewährleistet werden. Die bezüglich der Verfahr-
 richtung vorderen Quarzlichtleiter **4** nehmen die Remission des nicht gelaserten, untherapierten Hautareals auf, während die gegenüberliegenden, hinteren Quarzlichtleiter **4** die Remission des gelaserten, therapierten Hautareals registrieren. Durch eine geeignete Auswertung der Remissions- und anderen Sensorsignale kann daher bei der Gefäßtherapie eine un-
 gewollte Mehrfach-Laserung des Hautgewebes vermieden oder bei größeren Gefäßen eine Anzeigeeinrichtung ange-
 steuert werden, die dem Therapeuten bei der manuellen Bewegung des Laserhandstückes über der Haut einen Richtungs-
 hinweis liefert. Dazu sind auf der oberen Seite des Laserhandstückes elektronische Indikatoreinrichtungen **13** angebracht, die so angesteuert werden, dass die Fortbewegung in Richtung eines zu therapierenden Gefäßes durch einen leuchtenden LED-Pfeil gekennzeichnet wird. Dies ist nicht zuletzt deshalb notwendig, da die bevorzugte Verfahr-
 richtung des Laserhandstückes durch die Kontaktkühleinrichtung **8** vorgegeben ist. Diese liegt auf der Haut auf und kann die Lage der zu therapierenden Gefäße verdecken. Zur berührungslosen Bestimmung der Gewebetemperatur ist weiterhin eine Infrarot-
 strahlungs-Temperaturmesseinrichtung **5** vorgesehen, die aus einem Lichtleitermaterial gefertigt wird, welches sehr gut infrarote Strahlung bei 10 µm durchlässt. Gegenüber dem vorgenannten Kontaktthermometer **7** lassen sich hiermit Aus-
 sagen zum Temperaturverhalten des Hautgewebes ohne Druckbeeinflussung machen.

[0038] Die Messanordnung zur remissionsgesteuerten, selektiven Lasertherapie von Blutgefäßen und Hautgeweben ist in **Fig. 4**: dargestellt und besteht aus einem Lasergerät **14**, vorzugsweise aus einem Diodenlaser mit einer Wellenlänge von 810 nm, einer Frequenz bis 4 Hz, einer variablen Energiedichte bis 64 J/cm² und einer variablen Pulsbreite bis 60 ms. Das Lasergerät **14** ist in seinen Betriebsparametern wie Laserfluence, Pulsdauer und Pulsfrequenz durch die elektronische PC- oder Mikrorechner-Steuerungsbaugruppe **17** einstell- oder regelbar. Die Laserstrahlung wird durch eine
 Lichtleiteinrichtung **2** an das Laserhandstück mit einer Spotgröße bis zu 12 mm übertragen und tritt als eine Anzahl von Photonen **10** in das Hautgewebe **9** ein. Eine stabilisierte Wolfram-Halogen-Lampe mit einer Leistung bis zu 100 W dient als spektroskopische VIS-NIR-Strahlungsquelle **15**, deren Licht über ein Glasfaserbündel geleitet und die im Laserhand-
 stück als konzentrischer Ring ausgeführt wird. Sie kann durch die Steuerungsbaugruppe **17** geschaltet werden. Die Remissionssignale **12** werden in **Fig. 2** durch einen LOH-Quarzlichtleiter **4** aufgenommen und schematisch für alle konzen-
 trisch angeordneten Detektorlichtleiter (Anzahl n) an das VIS-NIR-Spektrometersystem **16** geführt. Im Eingang des VIS-NIR-Spektrometersystems **16** befindet sich entweder ein optischer Multiplexer, der nacheinander die n Detektor-
 lichtleiter **4** an ein VIS-NIR-Spektrometersystem im seriellen Betrieb anschaltet, oder es sind alternativ für einen Parallelbetrieb n VIS-NIR-Spektrometersysteme angeordnet. Das miniaturisierte VIS-Spektrometermodul besteht aus einem Glaskörper mit einem faseroptischen Eintritts-Querschnittswandler, einem abbildenden, holografisch geblazten Gitter als
 dispersivem Element sowie einer Silizium-Zeile in CCD- oder Diodentechnik. Die Silizium-Zeile weist eine Struktur von 256 Pixel auf, die im Wellenlängenbereich von 300 nm bis 1150 nm eine ausreichende Pixelauflösung von 3,2 nm
 gewährleisten. CCD-Zeilen haben einen größeren Dynamikbereich und ein geringeres Rauschen als Diodenzeilen. Die Silizium-Zeilen wandeln in bekannter Weise das spektral zerlegte Licht in elektrische Signale um, die über herkömmliche elektronische Schaltungen wie Verstärker, Multiplexer und Analog-Digital-Umsetzer der Auswerte- und Steuerungs-
 baugruppe **17** zugeführt werden. In konstruktiv ähnlicher Weise ist das miniaturisierte NIR-Spektrometermodul mit einer 128-er Diodenzeile aus InGaAs auf einem Quarzglaskörper aufgebaut. Für eine hohe Rauschunterdrückung ist hier eine integrierte thermoelektrische Kühlung auf 0°C erforderlich. Bei einem Wellenlängenbereich von 900 bis 1700 nm wird eine Pixelauflösung von 6,25 nm erreicht, die bekanntlich für die stark überlappenden NIR-Banden als ausreichend an-
 gesehen werden kann. Stromversorgungs- und Nachweisschaltungen der weiteren Sensoren **5**, **6**, **7** entsprechen dem be-
 kannten Stand der Technik und sind in **Fig. 4** nicht im einzelnen dargestellt. Die Auswerte- und Steuerungsbaugruppe **17**
 mit den jeweiligen Stromversorgungsbaugruppen besitzt einen PC-Card- oder Mikrorechnerkern, mit dem alle Ansteuerungs-
 aufgaben für die Spektrometerzeilen (Integrationszeit, Kompensation, Referenzierung), das Lasergerät und die Halogenlampe sowie die rechentechnische Auswertung der Remissions- und Sensorsignale vorgenommen werden. Dazu
 sind in der Auswerte- und Steuerungsbaugruppe **17** die entsprechenden remissionsspektroskopischen Modelle, Verfahren und Auswertelgorithmen gespeichert. Die zeitlichen Abläufe von Laserbestrahlung und Detektion durch die Mehr-
 fach-Sensorik werden durch die Steuerungsbaugruppe **17** derart realisiert, dass für herkömmliche Verfahrensgeschwindigkeiten im Bereich von 1 cm/s eine on line-Steuerung des Laserhandstück mit einem Mehrfach-Sensorsystem zur Diagnostik von Blutgefäßen und Hautgeweben ermöglicht wird.

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Liste der verwendeten Bezugszeichen

- 1 portables Gehäuse des Laserhandstückes
- 2 Laserstrahlungsübertragungseinrichtung
- 3 Halogenlampen-Strahlungsübertragungseinrichtung
- 4 Remissions-Strahlungsübertragungseinrichtung
- 5 Infrarotstrahlungs-Temperaturmesseinrichtung
- 6 Kontaktdrucksensoren
- 7 Kontakttemperatursensor
- 8 Kontaktkühleinrichtung
- 9 Hautgewebe
- 10 Laserstrahlung
- 11 Halogenlampenstrahlung
- 12 remittierte Photonen

13 elektronische Indikatoreinrichtungen

14 Lasergerät

15 VIS-NIR-Strahlungsquelle

16 VIS-NIR-Spektrometersystem

17 Auswerte- und Steuerungsbaugruppe sowie Stromversorgungsbaugruppe

5

Patentansprüche

1. Anordnung mit einem Laserhandstück zur sensorgesteuerten, selektiven Lasertherapie von Blutgefäßen und Hautgeweben sowie zur Steuerung von Parametern eines Lasertherapiegerätes **dadurch gekennzeichnet**, dass in unmittelbarer Nähe der Laserstrahlaustrittsfläche des Laserhandstückes in konzentrischer, polygonaler oder gegenüberliegender Form ein Mehrfach-Sensorsystem mit mindestens zwei gleichen oder unterschiedlichen Komponenten angeordnet ist. 10
2. Anordnung nach Anspruch 1, dadurch gekennzeichnet, dass mindestens eine Komponente des Mehrfach-Sensorsystems auf einem remissionsspektroskopischen Prinzip beruht und vorzugsweise mit sichtbaren und/oder naher infraroter Strahlung betrieben wird. 15
3. Anordnung nach Anspruch 2, dadurch gekennzeichnet, dass das Mehrfach-Sensorsystem die spektrale Remission von Blutgefäßen und Hautgeweben in berührender und/oder nichtberührender Weise über der Gewebeoberfläche aufnimmt.
4. Anordnung nach Anspruch 2, dadurch gekennzeichnet, dass das Mehrfach-Sensorsystem die spektrale Remission in nichtberührender Weise vor und nach dem zu therapierenden Gewebeareal aufnimmt, die konstruktiv durch eine geometrisch zurückgesetzte Anbringung der Lichtleiter der Remissionssensoren am Laserhandstück gewährleistet wird. 20
5. Anordnung nach Anspruch 1, dadurch gekennzeichnet, dass das Mehrfach-Sensorsystem Komponenten zum Nachweis der spektralen Remission sowie von Temperatur, Druck, Ultraschall und Feuchte enthält, die in unmittelbarer Nähe oder innerhalb der Laserstrahlaustrittsfläche vor, zwischen und nach Laserpulsen Sensorsignale über den Gewebezustand registrieren, woraus Steuerungssignale für ein manuelles oder automatisches Therapie regime des Lasergerätes abgeleitet werden. 25
6. Anordnung nach Anspruch 2, dadurch gekennzeichnet, dass die remissionsspektroskopische Komponente des Mehrfach-Sensorsystems zur lokalen Einschätzung des präoperativen Bräunungszustandes des Hautareals und weiterer physiologischer und morphologischer Gefäß- und Hautparameter wie der relativen Sauerstoffsättigung und dem relativen Wassergehalt verwendet wird. 30
7. Anordnung nach Anspruch 1 und 5, dadurch gekennzeichnet, dass eine Komponente des Mehrfach-Sensorsystems zur Temperaturmessung der Gewebeoberfläche dient, wodurch bei Überschreitung bekannter thermischer Grenzwerte eine Schädigung epidermaler und dermaler Bereiche durch Steuerung der Laserparameter vermieden wird. 35
8. Anordnung nach Anspruch 7, dadurch gekennzeichnet, dass zur Temperaturmessung nichtberührende Infrarot-Temperaturstrahlungsmesser und/oder berührende Halbleiter-Temperatur Sensoren/Thermoelementsensoren vorgesehen sind.
9. Anordnung nach Anspruch 1 und 3, dadurch gekennzeichnet, dass im Mehrfach-Sensorsystem bei berührender Betriebsweise eine Einrichtung zur Kontaktdruckmessung zwischen der Gewebeoberfläche und dem Handstück angeordnet ist, wodurch bei Über- bzw. Unterschreitung bekannter Druckgrenzwerte ein Warnsignal ausgelöst wird und eine unzulässige manuelle Gewebekompression bzw. ein Kontaktverlust vermieden werden kann. 40
10. Anordnung nach Anspruch 4, gekennzeichnet durch Verwendung von folgenden funktionsbestimmenden Komponenten: 45
 - einem gepulsten Hochleistungs-Diodenlaser mit rotem Zielstrahl laser, vorzugsweise beim Diodenlaser mit einer Wellenlänge von 810 nm, einer Frequenz bis 4 Hz, einer variablen Energiedichte bis 64 J/cm² und einer variablen Pulsbreite bis 60 ms
 - einem Laserhandstück und integrierter Kontaktkühlung
 - LED-Indikatoreinrichtung zur Anzeige der Verfahr richtung des Laserhandstückes 50
 - Lichtleiteinrichtungen zur Laserstrahlzuführung und Beleuchtung
 - LOH-Quarzlichtleiter zur VIS-NIR-Remissionsspektroskopie
 - einem oder mehreren VIS-NIR-Spektrometermodulen
 - einer Halogen glühlampe als spektroskopische Beleuchtungsquelle
 - einem IR-Lichtleiterbündel 55
 - einem integrierten Temperaturstrahlungsdetektor
 - einem integrierten Kontaktdrucksensor
 - Auswerte-Steuerungs-Elektronik (Build-In-Computer aus PC-Card- oder Mikrorechner-Baugruppen)
 - Stromversorgungsbaugruppen
11. Anordnung nach Anspruch 3, dadurch gekennzeichnet, dass zur Zuführung der Halogenlampen-Strahlung im Laserhandstück entweder die Laserstrahlungsübertragungseinrichtung oder ein separates LOH-Quarzlichtleiterbündel mit konzentrischer, polygonaler oder gegenüberliegender Anordnung der Lichtaustrittsseiten vorgesehen ist. 60
12. Anordnung nach Anspruch 11, dadurch gekennzeichnet, dass eine alternierende Zuführung der Laser- und Halogenlampestrahlung in das zu therapierende Gewebe über einen optischen Umschalter oder eine entsprechende mechanische Shuttereinrichtung vorgesehen ist. 65
13. Anordnung nach Anspruch 1, dadurch gekennzeichnet, dass den Mitteln zur sensorgesteuerten, selektiven Lasertherapie eine CCD-Videokamera mit telezentrischer Zoom-Optik oder ein Bildleiter-System für die Bilderfassung und Dokumentation mit anschließender Computerbildanalyse zugeordnet ist.

14. Anordnung nach Anspruch 2, dadurch gekennzeichnet, dass ermittelte Farbmerkmale zur Charakterisierung von lokal varikösen, bereits lasertherapierten oder gesunden Gefäßabschnitten zur Steuerung der Laserparameter verwendbar sind.

15. Anordnung nach Anspruch 2, dadurch gekennzeichnet, dass die spektrale Remission von n unterschiedlich lokalisierten Detektor-Lichtleitern im Laser-Handstück entweder an die gleiche Anzahl von n Spektrometermodulen zur gleichzeitigen, parallelen Auswertung oder über einen entsprechenden Lichtleiterumschalter mit n Eingängen und einem Ausgang nacheinander an ein Spektrometermodul zur aufeinanderfolgenden, seriellen Auswertung geleitet wird.

16. Anordnung nach Ansprüchen 12 und 14, dadurch gekennzeichnet, dass die spektrale Remission und/oder das Videobild die Lokalisation großer venöser Gefäße und die Beurteilung des Therapieerfolges über eine elektronische Anzeige ermöglicht und eine exakte manuelle oder automatische Führung des Laserhandstückes über dem Gewebe gestattet wird, ohne dass man bei einer visuellen Begutachtung gezwungen ist, seine Laser-Schutzbrille abzusetzen.

17. Anordnung nach Anspruch 16, dadurch gekennzeichnet, dass Indikatorelemente (typischerweise acht LED oder LCD) an der Oberseite des Laserhandstückes angeordnet sind, von denen ein Lokalisationssignal eines varikösen Gefäßes in der Gewebeumgebung angezeigt wird.

18. Anordnung nach Anspruch 17, dadurch gekennzeichnet, dass Indikatorsignale aus der spektralen Remission und von anderen Sensoren lageabhängig von einem varikösen Gefäß und der normalen Gewebeumgebung abgeleitet und zur Steuerung der Laserparameter verwendet werden.

19. Anordnung nach Anspruch 3 und 11, dadurch gekennzeichnet, dass im Laserhandstück auswechselbare Teile mit Bestrahlungs- und Detektor-Lichtleitern unterschiedlicher Lichteintritts- und -austrittsflächen (plan, sphärisch, parabel- und hyperbelförmig) vorgesehen sind, die einen angepassten Remissionsnachweis ausgewählter variköser Gefäße bezüglich Tiefe und Struktur verwirklichen.

20. Anordnung nach Anspruch 3, dadurch gekennzeichnet, dass das Laserhandstück bei nichtberührender Betriebsweise mit Hilfe von Abstandshalter, Abstandssensor und Anzeigevorrichtung in einer definierten Lage über dem zu therapierenden Gewebe positioniert wird.

21. Anordnung nach Anspruch 20, dadurch gekennzeichnet, dass der Abstandssensor vorzugsweise nichtinvasiv nach einem optischen Triangulations- oder einem remissionsspektroskopischen Prinzip wirkt.

22. Anordnung nach Anspruch 3, dadurch gekennzeichnet, dass das Laserhandstück bei nichtberührender Betriebsweise eine Vorrichtung zur Kühlung des lasertherapierten Hautareals mittels Kältespray enthält, welches solche Streu- und Absorptionseigenschaften aufweist, die eine remissionsspektroskopische Untersuchung nicht behindern.

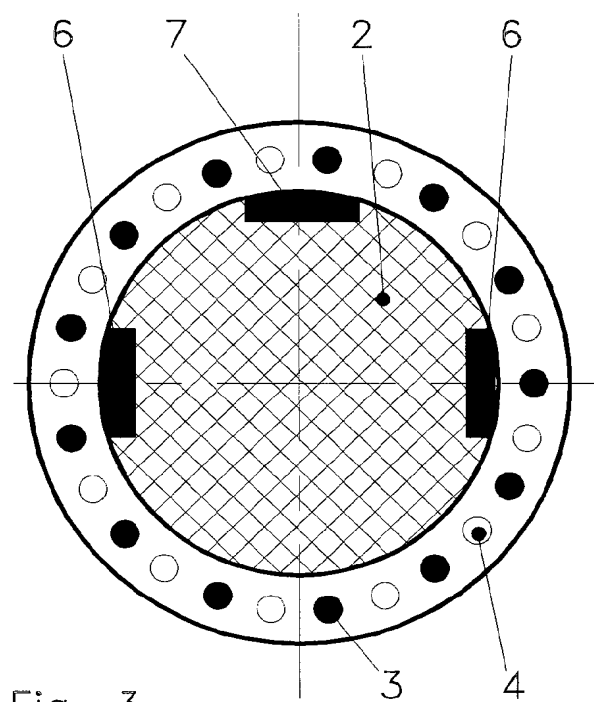
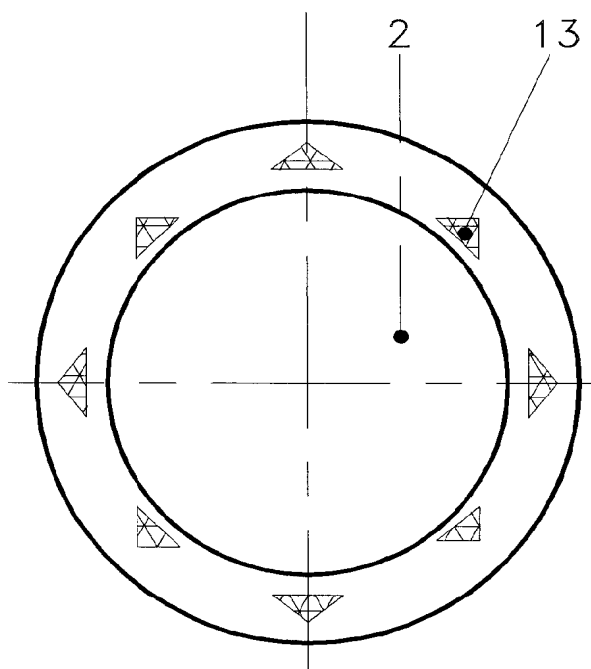
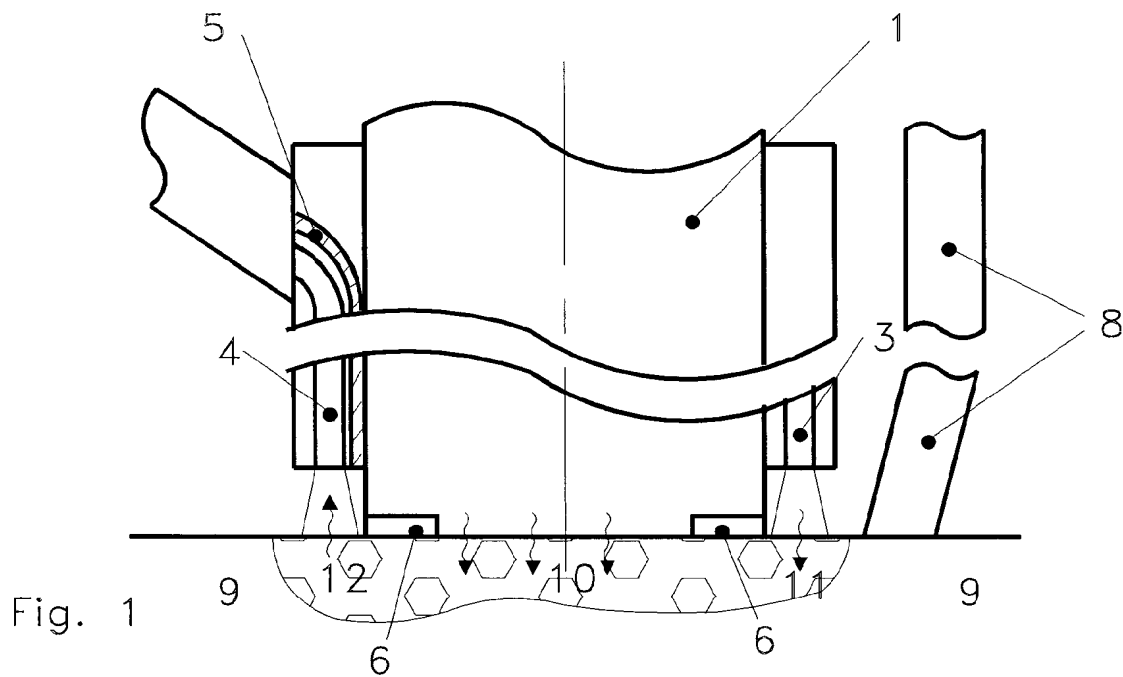
23. Anordnung nach Anspruch 1 und 5, dadurch gekennzeichnet, dass die Anordnung einen berührenden Ultraschallsensor enthält, der zur Lokalisation von Fettgewebe eingeschlossener, tieferliegender variköser Gefäße wirkt.

24. Anordnung nach Anspruch 1, gekennzeichnet durch die Verwendung zur schonenden Behandlung von Blutgefäßen, Wund- und Hautgeweben mit vergleichsweise niedrigen Laserenergien (Low Level Laser Therapie).

25. Anordnung nach Anspruch 24, dadurch gekennzeichnet, dass eine intra-orale oder epikutane Applikation von Dioden-Laserlicht am Nervenaustrittspunkt zur Beseitigung des vaskulären Kopfschmerzes mittels einer moderaten Trigemini-Reizung eingesetzt wird, wobei das Laserhandstück zur Lagelokalisation sowohl berührend als auch nichtberührend verwendbar ist.

26. Anordnung nach Anspruch 1, gekennzeichnet durch die Verwendung zur schonenden Entfernung von Tattoos und Schmutztätowierungen, wobei durch die vornehmlich spektroskopische Beurteilung der Pigmentzusammensetzung vor dem Lasereinsatz eine individuell maßgeschneiderte Therapie mit besseren Behandlungsaussichten und kalkulierbaren Nebenwirkungen ermöglicht wird.

Hierzu 2 Seite(n) Zeichnungen



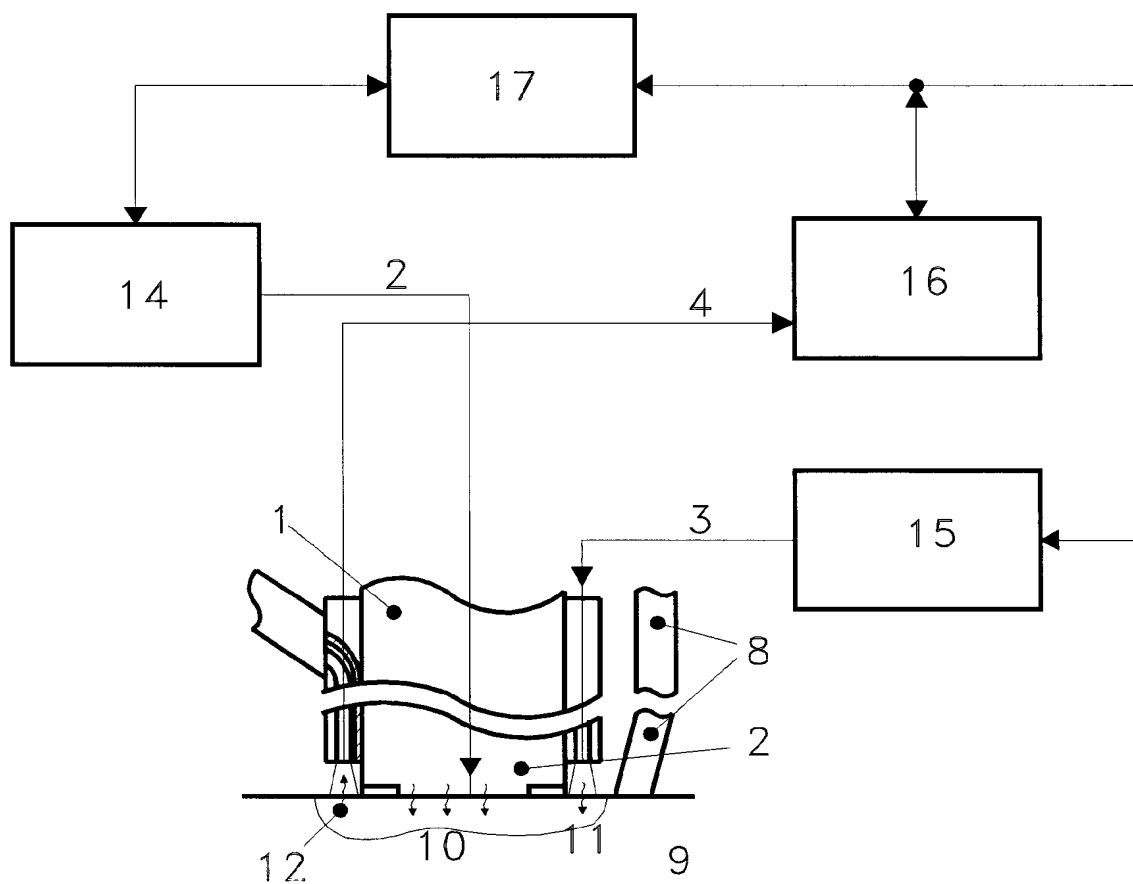





Fig. 4

Light therapy unit has infrared or polarizing filter in push in module is cooled by input air flow**Publication number:** DE10140715 (A1)**Publication date:** 2002-03-14**Inventor(s):** BOETTGER RALF FREDERIC [CH]**Applicant(s):** B & P AG ROSCHACHERBERG [CH]**Classification:****- international:** **A61N5/06; A61N5/073; A61N5/00; A61N5/06;** A61N5/00;
(IPC1-7): A61N5/06**- European:** A61N5/06C2**Application number:** DE20011040715 20010824**Priority number(s):** DE20011040715 20010824; DE20002014735U 20000825**Also published as:** DE20014735 (U1) CH695664 (A5) AT5177 (U1)**Abstract of DE 10140715 (A1)**

The light therapy unit outer housing (1) has the infrared or polarizing filter in a push in module (7) in front of the light mount (10) with light (2) and heat reflection filter (8) and cooled by the inflow of cooling air towards the fan (3).

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DEUTSCHLAND**



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PATENT- UND
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Die folgenden Angaben sind den vom Anmelder eingereichten Unterlagen entnommen

⑤④ Lichttherapiegerät

⑤⑦ Ein Lichttherapiegerät, aufweisend ein Gehäuse, das aus mindestens zwei Teilen zusammengesetzt ist, die durch eine lösbare Verbindung aneinander befestigt sind und zusammen einen Innenraum begrenzen, wobei in dem Innenraum eine Lichtquelle und ein Lüfter zur aktiven Führung von Kühlluft im Gehäuse angeordnet sind, wobei das Gehäuse mindestens je eine Lüftungsöffnung für den Zutritt von Kühlluft in das Gehäuse bzw. für den Austritt der Kühlluft aus dem Gehäuse aufweist, ist dadurch gekennzeichnet, daß das Gehäuse eine Aufnahme aufweist, über die entnehmbar ein Filterelement in den Innenraum des Gehäuses einschiebbar ist, ohne daß die Verbindung zwischen den Gehäuseteilen geöffnet werden muß, wobei ein Mittel vorhanden ist, das dafür ausgelegt ist, mit einem eingeschobenen Filterelement lösbar in Eingriff gebracht zu werden.

DE 101 40 715 A 1

[0001] Die Erfindung betrifft ein Lichttherapiegerät zur Erzeugung von Licht in einem bestimmten Wellenlängenbereich für die kosmetische und/oder medizinische Behandlung eines menschlichen und/oder eines tierischen Körpers.

[0002] Die sogenannte Lichttherapie, gemäß der Licht bestimmter Wellenlänge zur Bestrahlung der menschlichen (oder tierischen) Haut eingesetzt wird, um bestimmte physische oder psychische Effekte zu stimulieren, gewinnt im medizinischen und kosmetischen Bereich zunehmend an Bedeutung. Erfolge lassen sich insbesondere bei der Behandlung unreiner Haut, Akne, bei der Aktivierung des Stoffwechsels sowie der Regeneration beschädigter Hautpartien erzielen. Die Therapie hat sich jedoch auch im Bereich der Wundheilung, bei Ermüdungserscheinungen, Depressionen, Herpes, chronischen Schmerzen, Migräne, und auch zur Prävention der vorgenannten Indikationen sowie zur Triggerpunktbehandlung als äußerst vielversprechend erwiesen. Die Bestrahlung kann im wesentlichen auf der gesamten Körperoberfläche durchgeführt werden, so z. B. im Gesicht, auf dem Rücken, im Schulter-Nacken-Bereich usw. Zu den Vorteilen der Therapie zählt u. a. auch die Tatsache, daß so gut wie keine unerwünschten Nebenwirkungen auftreten. Die Lichttherapie eignet sich demzufolge auch sehr gut zur Selbstbehandlung des Patienten bei sich Zuhause.

[0003] Aus diesem Grunde wurden auf dem Markt zunehmend Geräte für die Lichttherapie entwickelt, die darauf abzielen, dem laienhaften Benutzer eine unkomplizierte und bequeme Handhabung des Geräts zu ermöglichen.

[0004] Ein derartiges Lichttherapiegerät, das dafür ausgelegt ist, während der Verwendung vom Benutzer in der Hand gehalten zu werden, ist in der EP 0 311 125 offenbart. Dieses bekannte Lichttherapiegerät weist eine Lichtquelle sowie einen sogenannten "Brewster-Polarisator" auf, der das von der Lichtquelle ausgesandte Licht in linear polarisiertes Licht umwandelt. Das Licht tritt durch eine Lichtfilterplatte hindurch aus dem Gerät aus. Die gesamte vorbeschriebene Lichtanordnung befindet sich in einem Innengehäuse, welches aus zwei röhrenförmigen Teilen gebildet wird, die miteinander einen definierten Winkel einschließen und deren äußeren Enden jeweils von der Lampe bzw. der Lichtfilterplatte luftdicht verschlossen werden. Das Innengehäuse befindet sich in einem äußeren Gehäuse, welches äußere Gehäuse zudem einen Entlüfter enthält. Dieser ist, vom inneren Gehäuse beabstandet, in einem Teil des äußeren Gehäuses untergebracht ist, welcher als Handgriff ausgebildet ist. Die zur Kühlung benötigte Luft tritt an einem Ende des äußeren Gehäuses in einen ringförmigen Spalt zwischen dem inneren und dem äußeren Gehäuse in letztgenanntes ein, verläuft dann entlang dem inneren Gehäuse und tritt hinter dem Lüfter aus dem als Handgriff ausgebildeten Teil des äußeren Gehäuses aus. Das innere Gehäuse dient bei der bekannten Konstruktion dazu, die optischen Elemente Lampe und Brewster-Polarisator gegen eindringenden Staub zu schützen.

[0005] Eine weitere Vorrichtung zur Stimulierung biologischer Prozesse mittels polarisierten Lichts ist in der DE 32 20 218 beschrieben. Auch die aus dieser Druckschrift bekannte Konstruktion weist im wesentlichen eine Lampe, einen Polarisator, der als Brewster-Polarisator, als Spiegelpolarisator, als Polaroidfilter, als Nicol'sches Prisma oder als eine andere Anordnung ausgebildet sein kann, sowie, gemäß einigen Ausführungsformen, einen Ventilator zur Kühlung auf. Es wird jedoch betont, daß ein solcher Ventilator nicht wesentlich ist und demzufolge weggelassen werden kann. Des weiteren ist es aus dieser Druckschrift bekannt, vor der Lampe einen Infrarotfilter anzuordnen, um

das Auftreten unerwünschter Wärmeeffekte auf der behandelten Hautpartie zu vermeiden. Schließlich umfaßt die Anordnung ein aus mehreren Linsen aufgebautes Ablenkssystem, um das Licht der Lampe parallel zur optischen Achse auszurichten und eine möglichst gleichmäßige räumliche Verteilung zu erreichen. Die genannten Komponenten sind in einem röhrenförmigen, länglichem Gehäuse untergebracht. Auf der Unterseite des Gehäuses kann ein Träger angebracht sein, durch den die bekannte Vorrichtung beispielsweise an einer Tischplatte festgeklemt werden kann.

[0006] Die bekannte Vorrichtung weist den Nachteil auf, daß die Filter entweder innerhalb des Gehäuses angebracht und deshalb nicht oder nur schwer zugänglich sind oder aber in einem separaten Aufsatz getrennt von den übrigen Komponenten außerhalb des Gehäuses untergebracht sind, sodaß keine Kühlung des Filters durch einen im Gehäuse befindlichen Lüfter möglich ist. Eine Kühlung ist jedoch unabdingbar, da die heute erhältlichen Polarisationsfilter nur begrenzten Temperaturen standhalten.

[0007] Ausgehend vom genannten Stand der Technik ist es eine Aufgabe der Erfindung, ein Lichttherapiegerät anzugeben, bei dem der Filter leicht zugänglich sein sollte, wobei er dennoch eine Kühlung durch einen Lüfter erfährt.

[0008] Erfindungsgemäß wird diese Aufgabe durch ein Lichttherapiegerät gelöst, welches ein Gehäuse aufweist, das aus mindestens zwei Teilen zusammengesetzt ist, die durch eine lösbare Verbindung aneinander befestigt sind und zusammen einen Innenraum begrenzen, wobei in dem Innenraum eine Lichtquelle und ein Lüfter zur aktiven Führung von Kühlluft im Gehäuse angeordnet sind, wobei das Gehäuse mindestens je eine Lüftungsöffnung für den Zutritt von Kühlluft in das Gehäuse bzw. für den Austritt der Kühlluft aus dem Gehäuse aufweist, wobei das Gehäuse eine Aufnahme aufweist, über die entnehmbar ein Filterelement in den Innenraum des Gehäuses einschiebbar ist, ohne daß die Verbindung zwischen den Gehäuseteilen geöffnet werden muß, wobei ein Mittel vorhanden ist, das dafür ausgelegt ist, mit einem eingeschobenen Filterelement lösbar in Eingriff gebracht zu werden.

[0009] Da das erfindungsgemäße Lichttherapiegerät eine Aufnahme für den Einschub eines Filterelements aufweist, ist es möglich, das Filterelement, wenn das Gerät nicht in Betrieb ist, dem Gehäuse zu entnehmen. Die Entnahme ist insbesondere deshalb problemlos, weil das Filterelement nicht Teil eines luftdichten Innengehäuses ist, in welchem Fall das Auswechseln des Filterelements, ohne daß Staub in das Innengehäuse eintritt, schwierig wäre. Es ist somit möglich, das herausgenommen Filterelement beispielsweise zu reinigen, zu ersetzen oder teilweise zu ersetzen oder gegebenenfalls, beispielsweise durch ein zusätzliches Filterelement, zu ergänzen und anschließend wieder in die Lichttherapiegerät zurückzusetzen. Zudem liegt es im Ermessen des Benutzers, das Filterelement nur dann in das Gerät einzusetzen, wenn der Lüfter eingeschaltet ist, so daß keine Möglichkeit besteht, daß sich bei ausgeschaltetem Lüfter eventuell Staubteilchen auf dem Filterelement absetzen können. Das Filterelement kann dann, wenn es nicht eingesetzt ist, beispielsweise in einer luftdichten Schutzhülle aufbewahrt werden.

[0010] Bei dem Filterelement kann es sich zum einen um einen Polarisationsfilter, um einen Farbfilter oder um eine Kombination aus einem Polarisationsfilter und einem Farbfilter handeln. Im letztgenannten Fall bilden beide Filter zusammen das Filterelement. Der oder die Filter können gemeinsam auf einer Halterung angebracht sein, welche in die Aufnahme einschiebbar ist. Es sollte deshalb beim Einsatz eines solchen Kombinationsfilters sichergestellt werden, daß zwischen den beiden Filtern keine Luftzutrittsmöglich-

keit besteht, um eine Absetzung von Staub zwischen den einzelnen Filtern zu verhindern. Dies kann beispielsweise durch eine luftdichte Steckverbindung gewährleistet werden.

[0011] Wenn das Filterelement in die Aufnahme eingeschoben ist, befindet es sich in den von den Gehäuseteilen umschlossenen Innenraum, welcher auch den Lüfter enthält. Das eingeschobene Filterelement wird demzufolge von dem vom Lüfter bewegten Kühlluftstrom erfaßt und gekühlt. Dies ist insbesondere dann wichtig, wenn es sich bei dem eingesetzten Filterelement um einen stark wärmeempfindlichen Polfilter handelt, der lediglich bis zu einer Temperatur von maximal 70°C stabil ist. Um eine optimale Kühlung des eingesetzten Filterelements zu erreichen, ist es günstig, wenn sich eine Lüftungsöffnung für den Zutritt von Kühlluft unmittelbar benachbart der Aufnahme für ein Filterelement im Gehäuse befindet. Hierdurch wird gewährleistet, daß ein möglichst großer Teil der Kühlluft mit den beiden Seiten des Filterelements in Kontakt kommt. Die Lüftungsöffnung kann beispielsweise in Form länglicher rechteckiger Schlitze ausgespart sein. Zur Verbesserung der Strömungsverhältnisse wird es jedoch vorgezogen, daß die beiden Teile der Lüftungsöffnung in Form langgestreckter Ellipsensegmente, d. h. mit abgerundeten Kanten, ausgebildet sind. Es sollte darauf geachtet werden, daß ein möglichst großer Teil jeder Seite der Filterplatte von der Kühlluft bestrichen wird. Vorzugsweise bestreicht der Kühlluftstrom mindestens etwa 80 bis 90% der Fläche jeder Seite, um eine optimale Kühlung und Entstaubung der Filterplatte zu gewährleisten. Vorzugsweise sollten alle Lüftungsöffnungen so ausgebildet sein, dass möglichst wenig Streulicht aus dem Gehäuse austreten kann.

[0012] Bei dem Mittel, das dafür ausgelegt ist, mit einem eingeschobenen Filterelement lösbar in Eingriff gebracht zu werden, kann es sich beispielsweise um einen Schnappmechanismus handeln, der leicht von Hand wieder gelöst werden kann, wenn das Filterelement dem Gehäuse erneut entnommen werden soll. Es ist jedoch auch möglich, das Mittel als einfache Steckverbindung auszulegen. Das Mittel kann mit dem Filterelement entweder direkt oder indirekt über dessen Halterung in Eingriff gebracht werden. Durch das Mittel wird sichergestellt, daß das Filterelement nicht ungewollt aus der Aufnahme herausrutscht, beispielsweise wenn die Leuchte gekippt wird.

[0013] Das zwei- oder mehrteilige Gehäuse des erfindungsgemäßen Lichttherapiegerätes ist vorzugsweise aus einem Kunststoffmaterial, insbesondere aus ABS oder Polycarbonat hergestellt. Dies hat die Vorteile, daß die Gehäuseroberfläche leicht gereinigt werden kann, das Gehäuse selbst ein nur geringes Gewicht aufweist und die Materialkosten vergleichsweise niedrig sind.

[0014] Gemäß einer besonders bevorzugten Ausführungsform ist eine optischen Linse im vorderen Teil des Gehäuses angebracht, durch die das von der Lichtquelle emittierte Licht zumindest teilweise aus dem Gehäuse austritt, wobei in den optischen Strahlengang zwischen Linse und Leuchtmittel eine Lochblende gesetzt ist. Die Blende, welche optional ist, begrenzt das Lichtbündel auf einen bestimmten Durchmesser, wobei der Blendendurchmesser durch die Linse abgebildet wird. Die Lochblende kann mit einem Wahrad ausgestattet sein, mittels dem der Lochblendendurchmesser mechanisch oder elektrisch verstellbar ist. Durch die Linse wird das von der Lichtquelle emittierte Licht gebündelt bzw. fokussiert, sodaß der Lichtstrahl gezielt auf die zu behandelnde Hautoberfläche gelenkt werden kann. Zudem sorgt die Linse dafür, daß der Mantel des Gehäuses, abgesehen von den notwendigen Lüftungsöffnungen, eine im wesentlichen geschlossene Fläche bildet, sodaß

der Eintritt von Staub und anderen kleinen Partikeln minimiert wird.

[0015] Aus Sicherheitsgründen ist es vorzuziehen, daß die Linse in einer Einstülpung in der Gehäusewand angebracht ist. Dies bedeutet, daß die Linse versenkt angeordnet ist. Auf diese Weise ist diese empfindliche optische Komponente gegen ein versehentliches Zerkratzen oder eine Beschädigung beim Herunterfallen des Gerätes besser geschützt. Neben dem Schutz vor einer Beschädigung bietet die abgesenkte Lage der Linse auch einen Schutz vor Fettablagerungen, welche beispielsweise durch Berührung mit der menschlichen Haut entstehen können.

[0016] Gemäß einer bevorzugten Ausführungsform des erfindungsgemäßen Lichttherapiegerätes ist die mindestens eine Lüftungsöffnung für den Zutritt von Kühlluft in das Innere des Gehäuses in einem Bereich des Gehäuses ausgespart, welcher eine gekrümmte Oberfläche aufweist. Diese Ausgestaltung hat den Vorteil, daß die mindestens eine Lüftungsöffnung auf diese Weise nicht dadurch blockiert bzw. versperrt werden kann, daß das Gerät im Bereich der Lüftungsöffnung auf der Auflagefläche aufliegt. Die mindestens eine Lüftungsöffnung kann demzufolge nicht durch ein falsches Ablegen der Leuchte von der Luftzufuhr abgeschnitten werden. Wie dies leicht vorstellbar ist, könnte eine Unterbrechung der Kühlluftzufuhr zu einer Überhitzung und damit Beschädigung einzelner Komponenten der Leuchte und insbesondere eines Filters führen. Zusätzlich kann natürlich auch ein Überhitzungsschutz vorhanden sein. Dieser kann beispielsweise aus einem Sensor bestehen, der die Temperatur an der empfindlichsten Stelle der Leuchte erfaßt und ein Temperatursignal an einen Regler ausgibt, welcher die Lichtquelle abschaltet, wenn das vom Sensor übertragene Signal einen vorbestimmten Schwellwert übersteigt. Die genaue Ausgestaltung eines solchen Notabschalters dem Fachmann auf dem Gebiet der Elektronik bekannt und soll hier deshalb nicht näher erläutert werden.

[0017] Die mindestens eine Lüftungsöffnung für den Zutritt von Kühlluft kann als Gruppe von Lüftungsöffnungen ausgebildet sein, die drei bis elf längliche Schlitzöffnungen umfaßt. Diese Öffnungen müssen nicht notwendigerweise dieselbe Länge und/oder Breite aufweisen. Falls sie jedoch bezüglich der Größe unterschiedlich ausgebildet sind, ist darauf zu achten, daß die zu beiden Seiten des genannten Zentrums ausgesparten Öffnungen der Gruppe bezüglich Größe und Form symmetrisch zueinander sind. Falls das Gehäuse im Bereich der Gruppe von weiteren Lüftungsöffnungen rotationsmetrisch ausgebildet ist, können sich die Öffnungen der Gruppe über einen Winkelbereich von bis zu 180° erstrecken. Die Gruppe von Schlitzöffnungen befindet sich vorzugsweise im vorderen Teil des Gehäuses, so daß der vom Lüfter angesaugte Luftstrom sowohl ein in die Aufnahme eingeschobenes Filterelement als auch die Lampe bestreicht. Der Luftstrom wandert somit durch einen Großteil des Innenraumes des Gehäuses.

[0018] Der Lüfter des erfindungsgemäßen Lichttherapiegerätes ist, von der Aufnahme für ein Filterelement aus gesehen, vorzugsweise hinter der Lichtquelle angeordnet, und die mindestens eine Lüftungsöffnung für den Austritt der Kühlluft aus dem Gehäuse befindet sich hinter dem Lüfter. Dies bedeutet, daß die Luftströmung von der mindestens einen Lüftungsöffnung für den Zutritt von Kühlluft weiter zur Lichtquelle und an dieser vorbeigeleitet wird, bis sie über den Lüfter durch die Austrittsöffnungen aus dem Gehäuse herausgeführt wird. Somit wird auch die Lichtquelle selbst durch den Kühlmittelstrom abgekühlt. Bei der Lichtquelle kann es sich beispielsweise um eine Kaltlichthalogenlampe, die sowohl für den Niedervolt- als auch für den Hochvoltbereich ausgelegt sein kann, handeln.

[0019] Vorzugsweise ist die mindestens eine Lüftungsöffnung für den Austritt der Kühlluft als Lüftungsgitter ausgebildet. Dies hat gegenüber einer oder wenigen größeren Luftaustrittsöffnung(en) den Vorteil, daß eine Beschädigung der im Inneren des Gehäuses befindlichen Elemente durch ein unerwünschtes Eindringen von Gegenständen durch die Lüftungsöffnung(en) weitestgehend vermieden wird. Das Gitter weist vorzugsweise eine Maschengröße von etwa 1 bis 3 mm auf.

[0020] Es hat sich aus Schutzgründen als vorteilhaft erwiesen, wenn das Lüftungsgitter in einer Vertiefung des Gehäuses angebracht ist.

[0021] Ein weiteres vorteilhaftes Merkmal, welches die Handhabung des erfindungsgemäßen Lichttherapiegerätes erleichtert, ist das Vorhandensein eines Hauptschalters zum Ein- und Ausschalten, welcher in einer konkaven Einstülpung der Gehäusewand angebracht ist. Da die Gehäusewand an der Stelle des Hauptschalters konkav ist, befindet sich der Hauptschalter in einer abgesenkten Lage. Er kann somit nicht dadurch versehentlich betätigt werden, daß das Lichttherapiegerät an der Stelle des Hauptschalters auf eine Auflagefläche abgestellt wird und der Hauptschalter durch das Aufliegen auf der Fläche umgelegt wird. Ein unwillentliches Einschalten des Gerätes wird somit erschwert.

[0022] Gemäß einer besonders bevorzugten Ausführungsform ist ein Bedienfeld mit mindestens einem Bedienelement an einer Seite der Gehäusewand angebracht ist, über das ein oder mehrere Parameter für die Steuerung der Leuchte einstellbar sind. Die Steuerung erfolgt vorzugsweise über an einer oder mehreren Leiterplatten aufgebrachten Elektronikschaltungen. Auf diese Weise wird die Benutzung der erfindungsgemäßen Lichttherapiegerät für den Benutzer komfortabler. Die einstellbaren Parameter können beispielsweise die Intensität der Beleuchtungsstärke, die Bestrahlungsdauer und den Abruf gespeicherter Behandlungsprogramme umfassen. Die abrufbaren Bestrahlungsprogramme können vom Hersteller der Leuchte einprogrammiert werden, so daß die Verwendung des Gerätes insbesondere für den medizinischen Laien erleichtert wird. Es ist des weiteren möglich, daß über die Bedienelemente die Öffnung einer Lochblende einstellbar ist, welche im Gehäuse angeordnet ist und den Lichtstrahldurchmesser begrenzt.

[0023] Das mindestens eine Bedienelement kann als Foliertaster ausgebildet sein. Auch dies dient der Benutzersicherheit, da derartige Taster nicht leicht aus Versehen betätigt werden können. Zudem sind sie aufgrund ihrer Abwischbarkeit aus hygienischen Gründen vorzuziehen.

[0024] Auf dem Bedienfeld kann daneben ein Anzeigeelement zur Anzeige der eingestellten Parameter vorhanden sein. Hierdurch werden dem Benutzer die wichtigsten Informationen bezüglich des aktuellen Zustands des Gerätes vermittelt. Das Anzeigeelement kann beispielsweise als LCD-Display ausgebildet sein.

[0025] Die aus dem Lichttherapiegerät austretende Kühlluft, die gegenüber der einströmenden Luft erwärmt ist, kann vorteilhaft dazu genutzt werden, aus einem Aromaspeicher Düfte freizusetzen. Ein solcher Aromaspeicher wird zweckmäßigerweise in der Nähe des Kühlluftaustrittes angebracht, entweder innerhalb des Gehäuses als Einschub, in einer ähnlichen Gestaltung wie beim Einschub des Filterelementes, oder außerhalb des Gehäuses als Aufsatz. Gegebenenfalls können an der Außenseite des Gehäuses Luftführungen vorgesehen sein, um den Duftstrom in Richtung auf die Vorderseite des Lichttherapiegerätes umzuleiten, diese sind jedoch nicht unbedingt erforderlich, da beispielsweise der sich selbst behandelnde Benutzer ohnehin zumindest in der Nähe des austretenden Duftstromes sein wird. Als Aromaspeicher kommen beispielsweise poröse Duftöle tra-

gende Materialien in Frage, die gegebenenfalls mit Luftdurchtritten versehen sind. Auch können Duftträger verwendet werden, die von der erwärmten Kühlluft umströmt werden. Die erwärmte Luft sorgt dafür, daß die Düfte in ausreichender Konzentration ausgetragen werden.

[0026] Auf dem Bedienfeld kann daneben ein Anzeigeelement zur Anzeige der eingestellten Parameter vorhanden sein. Hierdurch werden dem Benutzer die wichtigsten Informationen bezüglich des aktuellen Zustand des Gerätes vermittelt. Das Anzeigeelement kann beispielsweise als LCD-Display ausgebildet sein.

[0027] Weitere Merkmale und Vorteile der Erfindung gehen aus der nachfolgenden Beschreibung einer bevorzugten Ausführungsform derselben hervor, die als nicht beschränkendes Beispiel anhand der beigefügten Zeichnungen gegeben ist. In den Zeichnungen zeigen:

[0028] Fig. 1 ein erfindungsgemäßes Lichttherapiegerät im Längsschnitt;

[0029] Fig. 2 das Lichttherapiegerät von Fig. 1 im Querschnitt an der Stelle der Pfeile B-B gesehen;

[0030] Fig. 3 die gleiche Darstellung wie Fig. 1, wobei jedoch die ungefähren Strömungsverhältnisse eingezeichnet sind;

[0031] Fig. 4 eine Seitenansicht des erfindungsgemäßen Lichttherapiegerätes bei geschlossenem Gehäuse;

[0032] Fig. 5 eine Draufsicht auf das erfindungsgemäße Lichttherapiegerät;

[0033] Fig. 6 das erfindungsgemäße Lichttherapiegerät von vorne gesehen; und

[0034] Fig. 7 eine Längsschnittansicht durch das Gehäuse in der Ebene der Pfeile F-F gesehen.

[0035] In den Fig. 1 bis 7 ist eine Ansicht einer besonders bevorzugten Ausführungsform des erfindungsgemäßen Lichttherapiegerätes gezeigt, die von einem Gehäuse 1 umschlossen ist, welches beispielsweise aus einem geeigneten Kunststoff hergestellt sein kann. Bei der Auswahl des Kunststoffes ist darauf zu achten, daß er der durch die Strahlung entstehenden Temperaturbelastung standhalten kann. Insbesondere sollte er gegen lokal auftretende Temperaturgrenzwerte von bis zu 85°C beständig sein. Des weiteren ist eine bestimmte chemische Resistenz des eingesetzten Kunststoffes gegenüber den gängigen Desinfektionsmitteln sowie eine gewisse Biokompatibilität erforderlich, um eine hygienisch einwandfreie Benutzung des Gerätes gewährleisten zu können. Als geeignete Materialien haben sich insbesondere ABS oder Polycarbonat erwiesen.

[0036] Im Innenraum des Gehäuses 1 befinden sich die wesentlichen Elemente des erfindungsgemäßen Lichttherapiegerätes, nämlich insbesondere eine Lichtquelle 2 und ein Lüfter 3 zum Führen der Kühlluft innerhalb des Gehäuses. Bei der Lichtquelle kann es sich um eine Kaltlichthalogenlampe handeln, die im Niederspannungs- oder Hochspannungsbereich betrieben wird. Eine derartige Lampe emittiert etwa 12 bis 15% ihrer Strahlung im sichtbaren Bereich von etwa 400 nm bis etwa 780 nm. Die Strahlung ist somit für den Benutzer völlig ungefährlich. Als geeignet hat sich beispielsweise eine Halogenlampe mit einer Leistungsaufnahme von etwa 50 bis 100 Watt erwiesen. Derartige Halogenlampen weisen für gewöhnlich an ihrer Rückseite einen Reflektor 2a auf, der hier paraboloidförmig ausgebildet ist. Der Lüfter 3 kann von jeder bekannten geeigneten Art sein. Es kann sich beispielsweise um einen Flügelradlüfter handeln, der ähnlich einem Ventilator aufgebaut ist.

[0037] Zur Belüftung des Gehäuses sind verschiedene Lüftungsöffnungen 4, 5 und 6 im Gehäuse ausgespart, wobei die Lüftungsöffnungen 4 und 5 dem Zutritt von Kühlluft in das Gehäuse und die Lüftungsöffnungen 6, die hier als Verschlußgitter für den hinteren Teil des Gehäuses ausgebil-

det ist, dem Austritt der erwärmten Kühlluft nach hinten aus dem Gehäuse heraus dienen. Die erste Lüftungsöffnung **4** befindet sich an der Oberseite des Gehäuses **1** unmittelbar benachbart dem äußeren Rand einer Aufnahme **7** für ein Filterelement. Wie dies am besten in der Darstellung der **Fig. 4** und **5** zu sehen ist, ist die Lüftungsöffnung **4** im vorliegenden Fall zweiteilig ausgebildet, wobei sich die erste Hälfte **4a** vor der Aufnahme **7** und der zweite Teil der Lüftungsöffnung **4b** hinter dieser Aufnahme am Ende von Vertiefungen für die Entnahme eines eingesetzten Filterelements befinden. Die Teile **4a** und **4b** der Lüftungsöffnung **4** müssen nicht symmetrisch ausgebildet sein. Der Grund für diese besondere Ausgestaltung liegt in der Art der Luftführung, welche in der Darstellung von **Fig. 3** angedeutet ist. Zusätzlich zu der ersten Lüftungsöffnung **4** existiert bei der vorliegenden Ausführungsform nämlich noch eine weitere Lüftungsöffnung **5** für den Zutritt von Kühlluft, welche weitere Lüftungsöffnungen **5** hier als eine Gruppe von schlitzförmigen Öffnungen an der Unterseite des Gehäuses **1** ausgebildet ist. Der zusätzliche, durch die Lüftungsöffnung **5** eintretende Luftstrom trägt, wie dies in **Fig. 3** zu sehen ist, hauptsächlich zur Kühlung des hinteren Teils eines in die Aufnahme **7** eingeschobenen plattenförmigen Filterelements **7a** bei. Hingegen wird der vordere Teil des Filterelements kaum von dieser zusätzlichen Luftströmung erfaßt. Demzufolge ist es sinnvoll, den Teil **4a** der ersten Luftzutrittsöffnung **4** verschieden von dem zweiten Teil **4b** auszubilden.

[0038] Der durch die Öffnungen **4a** und **4b** eintretende Luftstrom verläuft zunächst entlang der Ebene des plattenförmigen Filterelements vor und hinter dem Filterelement, wobei sich der vor dem Filterelemente verlaufende erste Luftstrom an der Unterseite des Elements mit dem zweiten Luftstrom, welcher an der Öffnung **5** eintritt, vereinigt und mit diesem zusammen entlang dem hinteren Teil des Filterelements geführt wird. Die Luftströmung fließt dann in Richtung des rückwärtigen Endes des Gehäuses, wobei sie in der vorliegenden Ausführungsform auf einen Wärmereflektionsfilter **8** trifft, welcher vor der Lichtquelle angeordnet ist. Durch den Wärmereflektionsfilter wird der Infrarotanteil der von der Lichtquelle **2** emittierten und in Vorwärtsrichtung laufenden Strahlung weitestgehend ausgefiltert. Zusätzlich zur durch die Luftströmung bewirkten Konvektionskühlung findet auf diese Weise eine zusätzliche Kühlung durch Abschirmung von Wärmestrahlung statt.

[0039] Der Wärmereflektionsfilter **8** kann, wenn dies die Gegebenheiten zulassen, weggelassen werden. Falls er jedoch vorhanden ist, ist er, vorzugsweise mittels einer einfachen Schraubverbindung **9**, an einem Träger **10** für die Lichtquelle **2** befestigt. Bei dem Träger **10** handelt es sich vorzugsweise um einen speziell geformten Aluminiumträger, welcher am besten in der Darstellung von **Fig. 2** zu erkennen ist und der beispielsweise als Strangpreßprofil hergestellt sein kann. Aufgrund der hohen Wärmeleitfähigkeit von Aluminium wird ein großer Teil der von der Lichtquelle abgegebenen Wärmeenergie zum Lüfter **3** hin geführt, an dem der Träger **10**, vorzugsweise ebenfalls über eine Schraubverbindung, befestigt ist. Der Lüfter **3** selbst wird hier von einem weiteren Träger **11** gehalten, der an seinen Außenkanten mit der Gehäusewand in Eingriff steht. Diese Verbindung zwischen der Gehäusewand und dem Träger **11** ist vorzugsweise als eine Steck- oder Klemmverbindung ausgebildet, die sich automatisch mit dem Öffnen oder Schließen des Gehäuses löst bzw. schließt. Auf diese Weise entsteht ein entnehmbarer Einsatz, bestehend aus dem Träger **11** mit darauf angebrachtem Lüfter **3**, dem an dem Lüfter befestigten Lichtquellenträger **10** sowie der in den Träger **10** eingesetzten Lichtquelle **2** und gegebenenfalls mit dem ebenfalls am Träger **10** befestigten Wärmereflektionsfilter **8**.

[0040] Damit das Gehäuse **1** leicht geöffnet werden kann, so daß das Innere des Gehäuses zugänglich ist, ist dasselbe vorzugsweise aus einem Gehäuseoberteil **1a**, einem Gehäuseunterteil **1b** sowie einem, gegebenenfalls einstückigen, Vorderteil **1c** zusammengesetzt. Die einzelnen Teile können beispielsweise aus Spritzguß hergestellt sein.

[0041] Der Vorderteil **1c** des Gehäuses **1** ist von den beiden hinteren, als Halbschalen ausgebildeten Teilen **1a** und **1b** etwa an der Stelle der Aufnahme **7** für ein Filterelement getrennt. Er kann beispielsweise durch eine geeignete Klemm- oder Steckverbindung oder eine ähnliche, vorzugsweise lösbare Verbindung an den beiden hinteren Teilen **1a** und **1b** befestigt sein. Bei dem Filterelement kann es sich um einen Polarisationsfilter, z. B. um einen Polfilter, zum Ändern der Polarisationsrichtung des von der Lichtquelle ausgestrahlten Lichtes handeln. Somit strahlt die erfindungsgemäße Lichttherapiegerät zu einem hohen Anteil (etwa 95%) einheitlich polarisiertes Licht aus. Alternativ oder zusätzlich zu dem Polarisationsfilter kann das Filterelement einen Farbfilter enthalten, der bestimmte Wellenlängen des erzeugten Lichtes ausfiltert. Je nach gewünschtem Behandlungseffekt kann so eine bestimmte Farbe oder ein Farbbereich für das vom Gerät ausgestrahlte Licht ausgewählt werden. Es kann beispielsweise ein mehrteiliger Farbfilter eingesetzt werden. Ein solcher mehrteiliger Farbfilter könnte insbesondere einen Rot-, Orange-, Gelb-, Grün-, Türkis-, Blau- und Violettfilter umfassen.

[0042] Das plattenförmige Filterelement ist vorzugsweise entnehmbar in der Aufnahme **7** angebracht, um den oder die Filter gegebenenfalls reinigen oder austauschen zu können.

[0043] Zudem ist es auf diese Weise möglich, das Filterelement bei Nichtbenutzung des Gerätes zu entnehmen und separat aufzubewahren, um es vor Verschmutzung zu schützen, wenn sich der Lüfter nicht in Betrieb befindet.

[0044] Wie dies am besten in der Vorderansicht von **Fig. 6** zu sehen ist, befinden sich bei der gezeigten und beschriebenen Ausführungsform im vorderen Teil **1c** des Gehäuses **1**, auf der Unterseite desselben, insgesamt neun schlitzförmige Lüftungsöffnungen **5** für den Zutritt von Kühlluft in das Gehäuse **1**. Die Länge der Lüftungsöffnungen **5** nimmt, ausgehend von der größten, zentralen Lüftungsöffnung zu beiden Seiten nach außen kontinuierlich ab, wobei jeweils um gleiche Distanzen von der zentralen Lüftungsöffnung beabstandete Öffnungen die gleiche Größe aufweisen.

[0045] Im vorderen Teil **1c** des Gehäuses **1** befindet sich des weiteren eine optische Linse **12**, die die Austrittsöffnung für das von der Lichtquelle **2** erzeugten Lichts bildet. Die optische Linse **12** erfüllt zum einen die Aufgabe, das austretende Licht zu bündeln, und zum anderen bildet sie einen Verschuß für das Gehäuse **1**, so daß bei zusammengesetztem Gehäuse der Luftzutritt in den Innenraum nur über die dafür vorgesehenen Lüftungslöcher **4** und **5** erfolgt. Auf diese Weise läßt sich der Strömungsverlauf der Kühlluft bei vorgegebener Gehäuseform genau bestimmen, und die Gehäuseform kann vom Konstrukteur im Hinblick auf eine optimale Kühlluftströmung entwickelt werden. Der Durchmesser der Linse **12** kann zwischen etwa 30 mm und 60 mm, vorzugsweise etwa 50 mm, liegen. Bei der dargestellten Ausführungsform befindet sich die optische Linse **12** in einer dafür vorgesehenen Vertiefung **13** des vorderen Gehäuseteils, so daß sie den bestmöglichen Schutz gegen Beschädigung beim versehentlichen Herunterfallen des Gerätes erfährt. Auch ein Zerkratzen der Linse wird durch deren gegen den Innenraum versenkte Lage weitgehend vermieden.

[0046] Der Strahldurchmesser des austretenden Lichtes kann durch eine einstellbare Lochblende **14** in einem bestimmten Bereich eingestellt werden. Es hat sich ein solcher

Bereich als besonders geeignet erwiesen, mit dem der Strahldurchmesser in einem Abstand von etwa 15 cm von der Linse **12** entfernt zwischen etwa 10 cm und 20 cm beträgt. Das Einstellen des Lochblendendurchmessers kann beispielsweise manuell über ein Lochblendenwahlrad **15** erfolgen, das hier an der Gehäuseunterseite aus diesem heraus vorsteht.

[0047] Die hier gezeigte und beschriebene Lichttherapiegerät ist als Tischgerät ausgebildet. Mittels eines optionalen Anschlusses **16** kann sie daneben auch auf einem (hier nicht abgebildeten) Stativ angebracht werden, um insbesondere auf verschiedene Höhen einstellbar zu sein. Schließlich kann alternativ oder zusätzlich ein Handgriff (nicht gezeigt) vorhanden sein, um das Gerät bequem transportieren oder sogar während des Betriebes von Hand halten zu können.

[0048] Die für den Betrieb des erfindungsgemäßen Lichttherapiegerätes benötigte Elektronik befindet sich in der gezeigten Ausführungsform teilweise im oberen Gehäuseteil **1a** unmittelbar unterhalb der Gehäusewand und teilweise an der Lagerung des Lochblendenwahlrads. Sie ist hier in Form zweier Leiterplatten **17** bzw. **18** ausgebildet. Die auf der Leiterplatte **18** am Lochblendenwahlrad **15** angeordnete Leiterplatte kann z. B. Elemente zum Erfassen der aktuellen Winkellage der Wahlradscheibe enthalten, so daß der für die Behandlung eingestellte Lochblendendurchmesser gespeichert wird. Daneben kann diese Elektronik einen DC-Motor oder einen Verstellantrieb enthalten, um den Lochblendendurchmesser gegebenenfalls automatisch einstellen zu können.

[0049] Die auf der Leiterplatte **17** im oberen Gehäuseteil **1a** enthaltene Elektronik umfaßt vorzugsweise eine oder mehrere der folgenden Funktionen: Ansteuerung des Leuchtmittels (Intensität, Dauer der Bestrahlung), Anzeigen der Parameter (Lichtspotdurchmesser, Intensität, Dauer der Bestrahlung usw.), Überwachung der thermischen Sicherheit und Lüfteransteuerung. Mit anderen Worten kann diese Elektronik einen Regelmechanismus enthalten, mit dem eine Überlastung des Gerätes vermieden wird. Die für den Betrieb des erfindungsgemäßen Lichttherapiegerätes notwendige Spannungsversorgung wird hier über einen dafür vorgesehenen Anschluß **19** durch ein externes Netzteil (nicht gezeigt) sichergestellt. Gemäß einer alternativen Ausführungsform kann jedoch auch eine interne Spannungsversorgung über eine eingebaute Energiequelle, z. B. einen Akkumulator, verwirklicht werden. Neben dem elektrischen Anschluß **19** befindet sich hier der Hauptschalter **20** für das Ein- und Ausschalten des Gerätes.

[0050] Wie dies am besten in der Darstellung der **Fig. 5** zu sehen ist, befindet sich auf der Oberseite **1a** des Gehäuses ein spezielles Feld **21**, auf dem Bedienelemente **22** sowie ein Anzeigeelement **23**, das beispielsweise als LCD-Anzeige ausgebildet sein kann, angeordnet sind. Hier sind die Bedienelemente **22** in Form dreier Taster, z. B. Folientaster, ausgeführt, um die Behandlungsprogramme, die Intensität der Bestrahlungsstärke und die Bestrahlungs- bzw. Leuchtdauer einstellen zu können. Die Anzahl der Bedienelemente sowie die auswählbaren Funktionen können vom Fachmann je nach Bedarf modifiziert bzw. angepaßt werden und sollen an dieser Stelle deshalb nicht näher erläutert werden.

[0051] Je nach Art der integrierten Elektronik kann das Anzeigeelement **23** beispielsweise folgende Werte bzw. Parameter darstellen:

- Intensität in %, wobei die maximale Leistung gleich 100% ist;
- Spotdurchmesser in mm;
- Bestrahlungsdauer in min.
- über die Bedienelemente auswählbare Behandlungsprogramme.

[0052] In **Fig. 1** ist punktrichliert in schematischer Weise der außerhalb des Gehäuses liegende Aromaspeicher **30** gezeigt, der am hinteren Gehäuseende vor den Austrittsöffnungen **6** für die erwärmte Kühlluft liegt. Als Alternative kann vorgesehen sein, den Aromaspeicher **30** in dem freien Raum **31** vor den Austrittsöffnungen **6** anzuordnen. Der Aromaspeicher **30** kann dann, ähnlich wie das Filterelement **7a**, plattenförmig gestaltet werden und in eine Aufnahme einzuschieben sein. Gegebenenfalls können, je nach Ausführungsform, am Aromaspeicher oder am Gehäuse des Lichttherapiegerätes Luftführungen vorgesehen sein, um den mit Duftstoffen beladenen Luftstrom in eine gewünschte Richtung umzulenken.

[0053] Die in der vorstehenden Beschreibung, in der Zeichnung sowie in den Ansprüchen offenbarten Merkmale der Erfindung können sowohl einzeln als auch in beliebiger Kombination für die Verwirklichung der Erfindung wesentlich sein.

Patentansprüche

1. Lichttherapiegerät, aufweisend ein Gehäuse (**1**), das aus mindestens zwei Teilen (**1a**; **1b**) zusammengesetzt ist, die durch eine lösbare Verbindung aneinander befestigt sind und zusammen einen Innenraum begrenzen, wobei in dem Innenraum eine Lichtquelle (**2**) und ein Lüfter (**3**) zur aktiven Führung von Kühlluft im Gehäuse angeordnet sind, wobei das Gehäuse mindestens je eine Lüftungsöffnung (**4**, **5**; **6**) für den Zutritt von Kühlluft in das Gehäuse (**1**) bzw. für den Austritt der Kühlluft aus dem Gehäuse aufweist, **dadurch gekennzeichnet**, daß das Gehäuse (**1**) eine Aufnahme aufweist, über die entnehmbar ein Filterelement in den Innenraum des Gehäuses (**1**) einschiebbar ist, ohne daß die Verbindung zwischen den Gehäuseteilen (**1a**, **1b**) geöffnet werden muß, wobei ein Mittel vorhanden ist, das dafür ausgelegt ist, mit einem eingeschobenen Filterelement lösbar in Eingriff gebracht zu werden.
2. Lichttherapiegerät nach Anspruch 1, dadurch gekennzeichnet, daß das Gehäuse (**1**) aus einem Kunststoffmaterial hergestellt ist.
3. Lichttherapiegerät nach Anspruch 1 oder 2, dadurch gekennzeichnet, daß eine optischen Linse (**12**) im vorderen Teil des Gehäuses angebracht ist, durch die das von der Lichtquelle (**2**) emittierte Licht zumindest teilweise aus dem Gehäuse austritt, wobei eine Lochblende mit einstellbarem Durchmesser in den optischen Gang zwischen die optische Linse und die Lichtquelle gesetzt ist.
4. Lichttherapiegerät nach einem der Ansprüche 3, dadurch gekennzeichnet, daß die Linse (**12**) in einer Einstülpung in der Gehäusewand angebracht ist.
5. Lichttherapiegerät nach einem der Ansprüche 1 bis 4, dadurch gekennzeichnet, daß die mindestens eine Lüftungsöffnung (**5**) für den Zutritt von Kühlluft in das Innere des Gehäuses in einem Bereich des Gehäuses (**1**) ausgespart ist, welcher eine gekrümmte Oberfläche aufweist.
6. Lichttherapiegerät nach einem der vorstehenden Ansprüche 1 bis 5, dadurch gekennzeichnet, daß die mindestens eine Lüftungsöffnung (**5**) für den Zutritt von Kühlluft als Gruppe von Lüftungsöffnungen ausgebildet ist, die drei bis elf längliche Schlitzöffnungen umfaßt.
7. Lichttherapiegerät nach einem der vorstehenden Ansprüche, dadurch gekennzeichnet, daß die mindestens eine Lüftungsöffnung (**6**) für den Austritt der

Kühlluft als Lüftungsgitter ausgebildet ist.

8. Lichttherapiegerät nach Anspruch 7, dadurch gekennzeichnet, daß das Lüftungsgitter in einer Vertiefung des Gehäuses angebracht ist.

9. Lichttherapiegerät nach mindestens einem der vorstehenden Ansprüche, dadurch gekennzeichnet, daß sie einen Hauptschalter **(20)** zum Ein- und Ausschalten aufweist, welcher in einer konkaven Einstülpung der Gehäusewand angebracht ist. 5

10. Lichttherapiegerät nach einem der Ansprüche 1 bis 9, dadurch gekennzeichnet, daß ein Bedienfeld **(21)** mit mindestens einem Bedienelement **(22)** an einer Seite der Gehäusewand angebracht ist, über das ein oder mehrere Parameter für die Steuerung der Leuchte eingegbar sind. 10

11. Lichttherapiegerät nach Anspruch 10, dadurch gekennzeichnet, daß die eingegbaren Parameter die Intensität der Bestrahlungsstärke, die Bestrahlungsdauer und den Abruf gespeicherter Behandlungsprogramme umfassen. 15 20

12. Lichttherapiegerät nach einem der Ansprüche 10 bis 11, dadurch gekennzeichnet, daß das mindestens eine Bedienelement **(22)** als Folientaster ausgebildet ist.

13. Lichttherapiegerät nach Anspruch 10, dadurch gekennzeichnet, auf dem Bedienfeld ein Anzeigeelement **(23)** zur Anzeige der eingestellten Parameter vorhanden ist. 25

14. Lichttherapiegerät nach einem der Ansprüche 1 bis 13, dadurch gekennzeichnet, daß im Bereich der aus dem Gehäuse austretenden Kühlluft ein Aromaspeicher **(30)** angeordnet ist. 30

15. Lichttherapiegerät nach einem der Ansprüche 1 bis 13, dadurch gekennzeichnet, daß ein Aromaspeicher in einem freien Raum **(31)** zwischen Lichtquelle **(2)** und Austrittsöffnungen **(6)** für die Kühlluft angeordnet ist. 35

16. Lichttherapiegerät nach Anspruch 14 oder 15, dadurch gekennzeichnet, daß am Gehäuse oder am Aromaspeicher **(30)** Luftführungen zum Umlenken des mit Duftstoffen beladenen Luftstromes vorgesehen sind. 40

Hierzu 4 Seite(n) Zeichnungen

45

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- Leerseite -

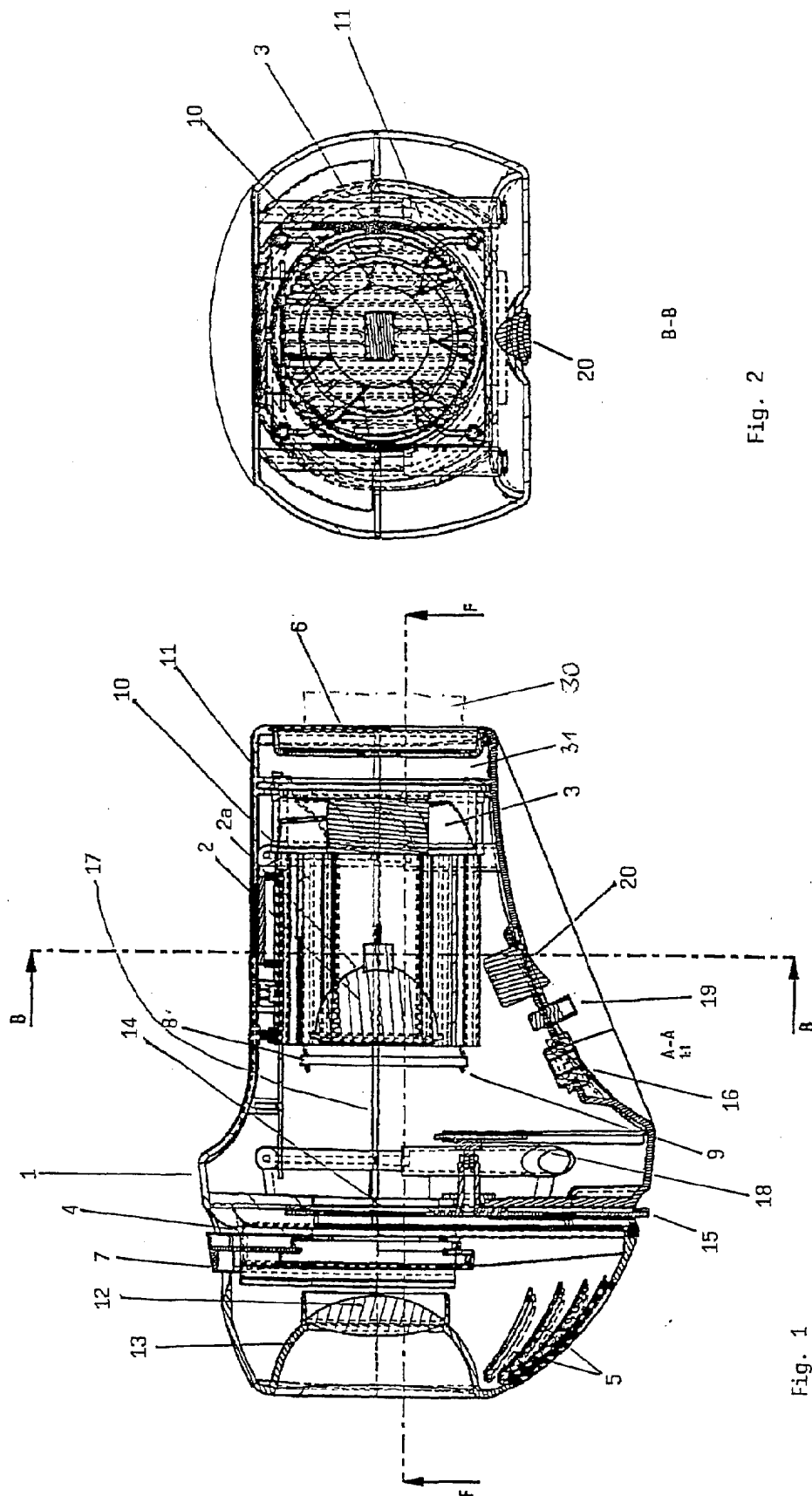
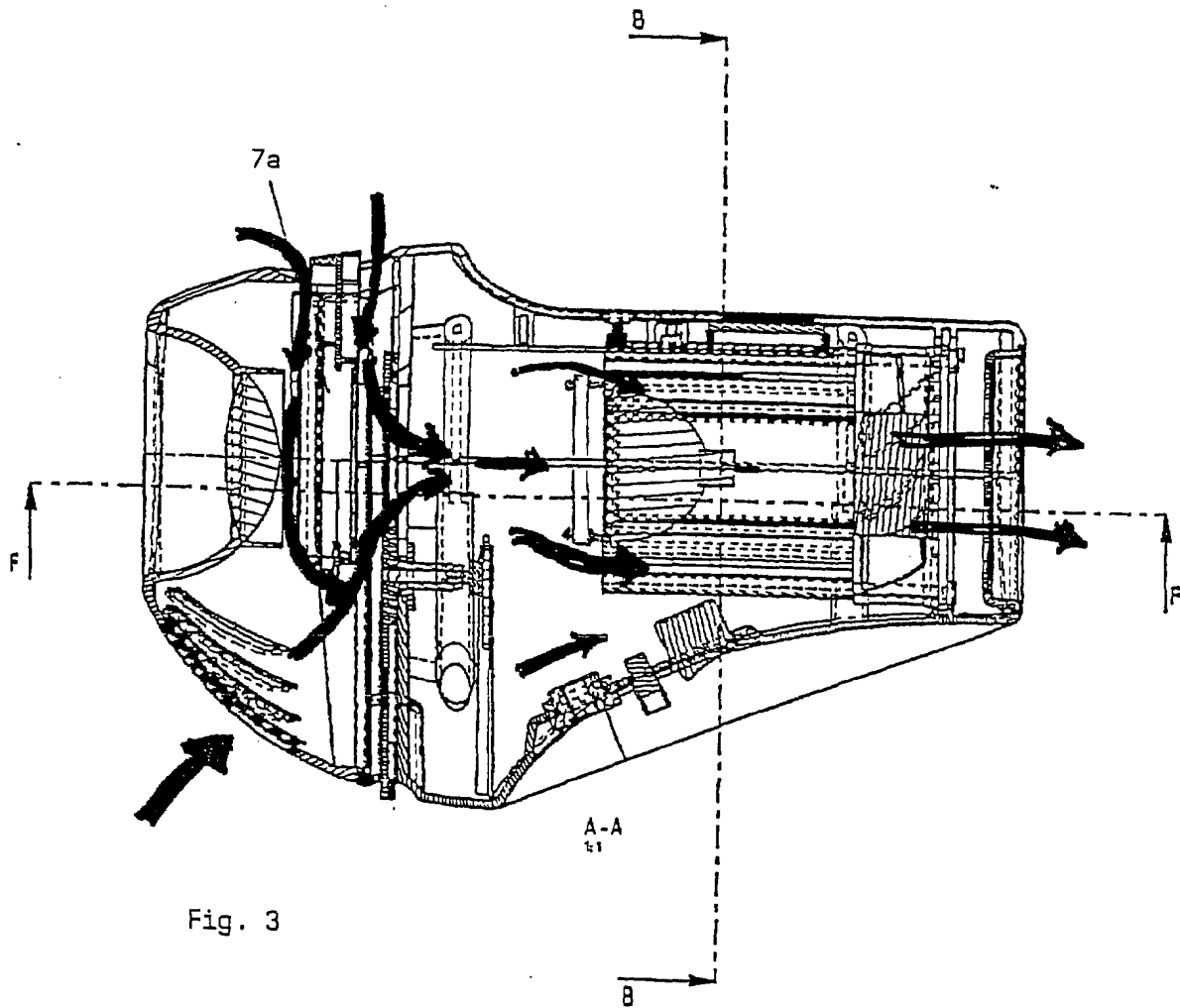


Fig. 2

Fig. 1



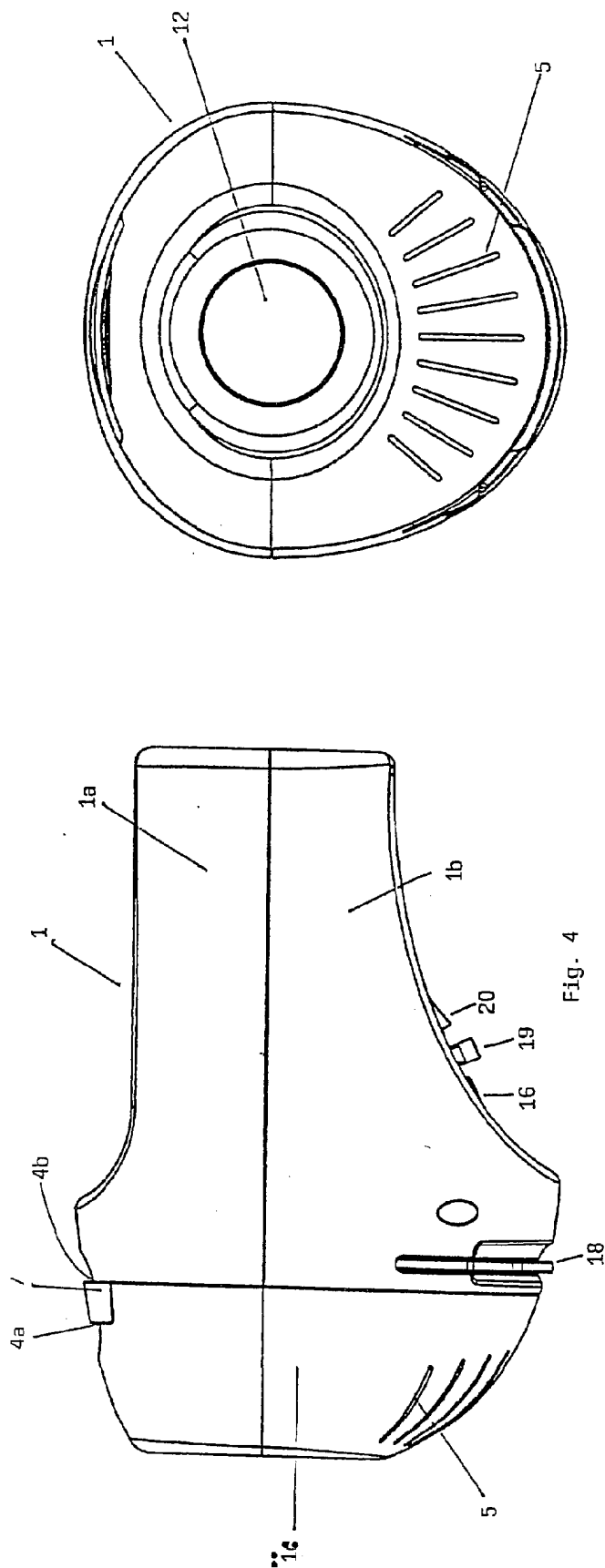


Fig. 4

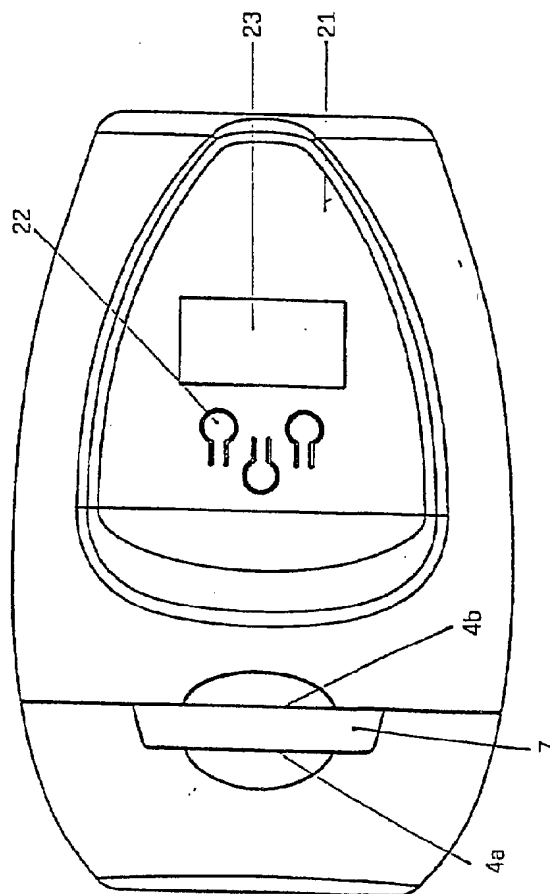
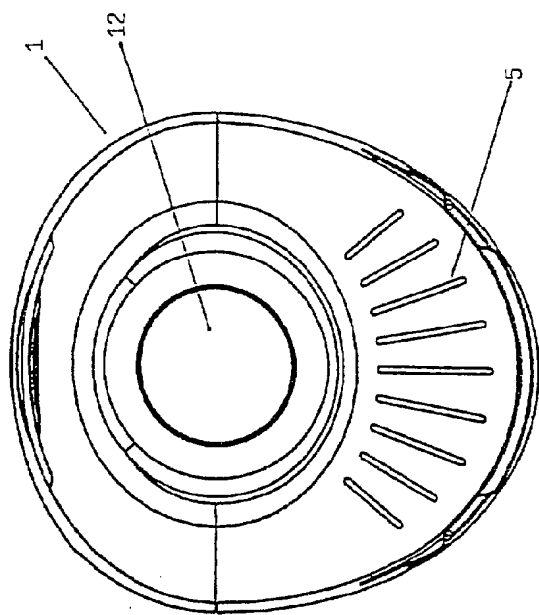


Fig. 5

Fig. 6



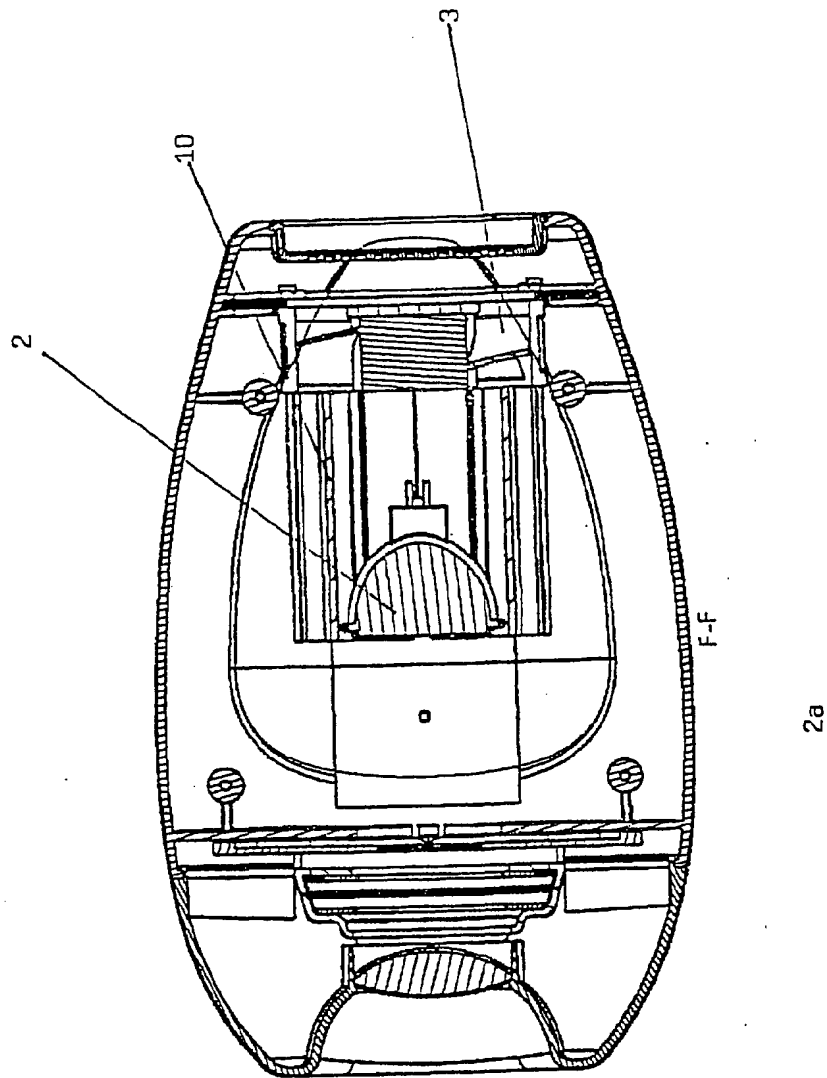


Fig. 7

Depth/structure-selective biological tissue treatment method and device e.g. for body hair removal, uses simultaneous application of pressure and irradiation with light

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Publication date: 2001-03-22

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Priority number(s): DE19991044401 19990916

Abstract of DE 19944401 (A1)

The treatment method and device has the tissue area to be treated subjected to an applied pressure during treatment with light of selected frequency, e.g. laser light, with cooling of the tissue surface and selection of the refractive index difference between the tissue surface and the environment. The pressure can be provided by a surface loading force or a suction force applied via a suction cup.

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⑮ **BUNDESREPUBLIK
DEUTSCHLAND**



**DEUTSCHES
PATENT- UND
MARKENAMT**

⑫ **Offenlegungsschrift**
⑩ **DE 199 44 401 A 1**

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Die folgenden Angaben sind den vom Anmelder eingereichten Unterlagen entnommen

⑤④ **Verfahren und Vorrichtung zur Struktur-/tiefenselektiven Denaturierung biologischen Gewebes**

⑤⑦ Die Erfindung betrifft ein Verfahren und eine Vorrichtung, die in der Lage sind, im Bereich oberflächennaher Gewebeschichten tiefen-/strukturselektive Denaturierung zu verursachen. Speziell soll es mit diesem Verfahren und der zugehörigen Vorrichtung möglich sein, Haarwurzeln in der menschlichen Oberhaut zu veröden, indem durch die Wahl geeigneter Parameter Wellenlänge, Bestrahlungsfläche, mechanische Kompression und Oberflächenkühlung beispielsweise eine thermische Denaturierung in einer Schichttiefe von 300 ± 50 bis 100 μm erfolgt. Dabei wird durch die Intensitätsüberhöhung des Streulichts in dieser Tiefe ein Temperaturprofil eingestellt, das in seinem Maximum eine Temperatur von deutlich über 57°C erreicht, so dass es innerhalb weniger Sekunden zu einer nachhaltigen Schädigung/Denaturierung des so thermisch gestressten Gewebes kommt.

DE 199 44 401 A 1

Beschreibung

Aufgabenstellung

Es sollen ein Verfahren und eine Vorrichtung gefunden und entwickelt werden, die in der Lage sind, im Bereich oberflächennaher Gewebeschichten tiefen-/strukturselektive Denaturierungen zu verursachen. Speziell soll es mit diesem Verfahren und der zugehörigen Vorrichtung möglich sein, Haarwurzeln in der menschlichen Oberhaut zu veröden.

Stand der Technik

Konventionelle Methoden zur Epilation (Rasieren der Haare, mechanische Epilation mittels Pinzetten, Epilation mit Enthaarungscremes oder Wachsepilation) führen in der Regel zu keinem lang anhaltenden Erfolg und können zu zahlreichen Nebenwirkungen und Komplikationen in Form von Hautentzündungen, Narbenbildung sowie Hyper- bzw. Hypopigmentierungen führen. Die Elektroepilation ist hingegen wirksamer, aber auch diese Methode erfordert eine mühsame, langwierige und aufwendige Prozedur, bei deren Anwendung das Auftreten von Komplikationen nicht selten ist. In der modernen Epilation werden folgende Effekte angewendet: Galvanische Elektrolyse, Photothermische Effekte (Koagulation, Erwärmung), Photo-mechanische Effekte (Schockwellen, Kavitation), Photochemische Effekte z. B. Photodynamische Therapie (PDT).

Das breite Spektrum der Indikationen sowie die Problematik, die sich aus der Anwendung der konventionellen Methoden zur Haarentfernung ergibt, gebietet die Suche nach neuen, verbesserten Systemen zur Epilation, deren Anwendung eine dauerhafte Haarentfernung gewährleisten kann.

Erfindungsgemäße Lösung

Überraschenderweise hat sich gezeigt, dass es durch die Streuung des Lichtes im biologischen Gewebe, obwohl diese Streuung eine extrem starke Vorwärtsstreuung charakteristisch hat, zu einer im zeitlichen Mittel signifikanten Intensitätsüberhöhung durch Vielfachstreuungsprozesse im Bereich oberflächennaher Strukturen kommt. Die Tiefenausdehnung dieser Intensitätsüberhöhung kann dabei erfindungsgemäß durch die Auswahl einer geeigneten Wellenlänge des zur Bearbeitung/Behandlung benutzten Lichtes eingestellt werden. Darüber hinaus wird die Tiefenlokalisation und -ausdehnung erfindungsgemäß durch die Größe des beleuchteten Oberflächenareals bestimmt und weiterhin durch die mechanische Kompression der Gewebeschichten beeinflusst. Die mechanische Kompression der Gewebeschichten führt dabei zu Strukturveränderungen und hat wesentlichen Einfluss auf die Streulichtverteilung des benutzten Lichtes. Gleichzeitig hat sich auch für den Fachmann völlig überraschend ergeben, dass bei geeigneter Wahl der Größe der beleuchteten Oberfläche die Tiefenausdehnung der Streulichtverteilung in weiten Grenzen eingestellt werden kann. Darüber hinaus kann die Tiefenselektion und -ausdehnung erfindungsgemäß durch eine geeignete, oberflächlich aufgetragene klare Flüssigkeit beeinflusst werden, indem eine Anpassung des Brechungsindex von Flüssigkeit und Gewebe erfolgt. Es zeigte sich überraschenderweise, daß durch Variation des Brechungsindex das Tiefenprofil der Intensitätsverteilung leicht beeinflusst werden kann. In Weiterführung des Erfindungsgedankens wird die zu bestrahlende Gewebeoberfläche durch geeignete Maßnahmen gekühlt. In einem besonders bevorzugten Ausführungsbeispiel wird ein für die benutzte Wellenlänge transparenter Eiskwürfel bzw. eine transparente Eisplatte auf die Oberflä-

che gelegt, so dass es im Gewebekontakt auf Grund des vorhandenen Temperaturgradienten zu einer Verflüssigung des Eises kommt und damit der Umgebung zusätzlich zu der durch Wärmeleitung bedingten Kühlung die Phasenumwandlungsenergie Eis zu Flüssigkeit entzogen wird.

Gleichermaßen kann statt eines Eiskwürfels mit einer flüssigen Grenzschicht zum Gewebe auch ein geeigneter aerosolgetragener Feuchtheitsfilm auf die Grenzfläche eingebracht werden, dergestalt, dass bereits bei der vorhandenen Grenzschichttemperatur bzw. bei leichter, durch den Prozess bedingter Temperaturerhöhung eine Verdampfung auftritt, wobei wiederum zusätzlich zur Wärmeleitung der Gewebegrenzschicht die Phasenumwandlungsenergie zur Verdampfung entzogen wird.

Im Rahmen eines bevorzugten Anwendungsbeispiels wird das Verfahren zur Denaturierung von Haarwurzeln genutzt, wobei erfindungswesentlich durch Wahl der geeigneten eingangs genannten Parameter Wellenlänge, Bestrahlungsfläche, mechanische Kompression und Oberflächenkühlung beispielsweise eine thermische Denaturierung in einer Schichttiefe von 300 ± 50 – $100 \mu\text{m}$ erfolgt. Dabei wird durch die Intensitätsüberhöhung des Streulichts in dieser Tiefe ein Temperaturprofil eingestellt, das in seinem Maximum eine Temperatur von deutlich über 57°C erreicht, so dass es innerhalb weniger Sekunden zu einer nachhaltigen Schädigung/Denaturierung des so thermisch gestressten Gewebes kommt.

In Weiterführung dieses Erfindungsgedankens ist eine lokale Erhöhung des Energieeintrages nicht nur durch Erhöhung der mittleren Streulichtintensität in der vorgesehenen Wirktiefe erfindungswesentlich, sondern zusätzlich werden beispielhaft zur Denaturierung der Haarwurzeln geeignete Farbstoffe, vorzugsweise Lebensmittelfarbstoffe, eingesetzt, die sich über die natürlicherweise vorhandenen porigen Hautöffnungen bzw. Follikel selektiv in den Haarwurzeln und anderen porigen Vertiefungen anreichern. Bei geeigneter Wahl dieser Farbstoffpigmente und der benutzten Laserstrahlung findet dann eine erhöhte Absorption des verwendeten Wirklichtes in eben dieser Tiefe statt, so dass insgesamt die zur Behandlung gewählte Flächenhelligkeit weiter reduziert werden kann und damit eine effektivere Schonung der oberflächlichen und unter der Behandlungszone liegenden tieferen Gewebeschichten erreicht wird. Unbenommen dieser technischen Maßnahmen zur Begrenzung der Denaturierungszone auf eine definierte Gewebstiefe wird allein durch die biologische Reaktion des thermisch induzierten Ödems und der dadurch bedingten Verlegung von Versorgungskanälen die natürliche Versorgung der darüber liegenden Zellschichten behindert bzw. unterbunden. Die Folge wäre ein zeitlich versetztes Absterben eben dieser oberflächlichen Zellschichten durch Versorgungsmangel. Erfindungsgemäß wird dieses dadurch verhindert, dass die bestrahlten Flächenareale in einer schachbrettartigen Anordnung bzw. in einer hexagonalen Kreisflächenstruktur so gewählt wird, dass bei Behandlung jeweils der übernächsten Nachbarfelder durch die nichtbehandelten dazwischenliegenden Felder hinreichend viel Nährsubstanz durch laterale Diffusion in die über der behandelten Zone liegenden Schichten gelangen kann und damit eine Vitalfunktion dieser Zellschichten erhalten bleibt.

Nach Ausheilen der Denaturierungsnekrose und Abklingen des dadurch bedingten Schwellungszustands (Ödem) findet dann auch wieder eine transversale Versorgung dieser Zellschichten statt, so dass dann in einem zweiten Verfahrensschritt die bisher ausgesparten intermediären Flächenfelder behandelt werden können.

Abb. 1 zeigt dabei die tiefenbegrenzte Intensitätsüberhöhung in Abhängigkeit von der benutzten Wellenlänge und

der bestrahlten Fläche. Bei der Auswahl einer Wellenlänge λ_1 (z. B. 585 nm) und einer bestrahlten Fläche A_1 (z. B. \varnothing 3 mm) resultiert aufgrund der ausgeprägten Gewebestreuung eine Intensitätsüberhöhung unmittelbar unter der Oberfläche des bestrahlten Areal. Vorzugsweise befindet sich das Maximum der Intensität im Bereich der zu zerstörenden biologischen Strukturen, d. h. in der Tiefe 300 μ m. Die Intensität der Strahlung im Maximum kann dabei ein Vielfaches der von außen applizierten Bestrahlungsstärke betragen. Der Effekt beruht auf der in biologischem Gewebe vorliegenden Vielfachstreuung, so daß ein einzelnes Lichtquant (Photon) ein Volumenelement mehrfach durchqueren kann und so zu einer relativen Intensität größer 1 führt. Der Intensitätsabfall direkt unter der Gewebeoberfläche beruht auf der Tatsache, daß Lichtquanten das Gewebe an der Grenzschicht wieder verlassen können. In tieferen Schichten fällt die Intensität entsprechend eines exponentiellen Schwächungsgesetzes ab. Wird nun bei gleicher Bestrahlungsgeometrie eine andere Wellenlänge λ_1 verwendet (z. B. 685 nm) verändert sich die Lage des Intensitätsmaximums in der Gewebetiefe, weil die optischen Gewebeparameter wellenlängenabhängig sind. Damit kann die Lage des Intensitätsmaximums der Lage der Schicht angepaßt werden, in der sich die zu zerstörenden biologischen Strukturen befinden. Eine weitere Beeinflussungsmöglichkeit besteht über den Durchmesser der zu bestrahlenden Fläche. Wird eine größere Fläche A_2 bestrahlt, kann die Anzahl der seitlich für den Prozeß verloren gegangenen Lichtquanten reduziert werden, so daß sich ein verändertes Tiefenprofil ergibt. In diesem Fall sind zusätzlich die Laserparameter anzupassen, um z. B. die Bestrahlungsstärke konstant zu halten.

Abb. 2 zeigt das Tiefenprofil der Temperaturverteilung im biologischen Gewebe, welches aus einer in **Abb. 1** gezeigten Kombination von Wellenlänge λ_1 und Bestrahlungsdurchmesser A_1 resultiert, bei der das Intensitätsmaximum in einer Tiefe von ca. 300 μ m liegt. Das Temperaturprofil ist bei kurzen Bestrahlungszeiten direkt mit der Intensitätsverteilung vergleichbar, durch Wärmeleitprozesse wird der Kurvenverlauf jedoch etwas abgeflacht.

Abb. 3 zeigt den Effekt, der durch eine strukturbegrenzte Erhöhung des Absorptionskoeffizienten verursacht wird. Es kommt im Bereich hoher Absorberkonzentrationen bei vergleichbarer Intensitätsverteilung zu einer vermehrten Absorption und in Folge zu einer ausgeprägteren Temperaturerhöhung. Durch die Selektivität der Absorptionserhöhung bleibt der Effekt bei kurzen Expositionszeiten weitestgehend auf die zu zerstörenden Strukturen begrenzt und erhöht somit die Effizienz der Methode.

Abb. 4a zeigt ein typisches schachbrettartiges Bestrahlungsmuster, um die postoperative irreversible Schädigung oberflächennaher Strukturen durch Versorgungsmangel zu verhindern. Die Größe der versetzt angeordneten bestrahlten Areale ist so zu wählen, daß durch laterale Diffusionsprozesse eine ausreichende Versorgung der oberflächlichen, zuvor thermisch nicht geschädigten Strukturen gewährleistet ist. Nach dem Abklingen des postoperativen Ödems kann die Behandlung der zuvor ausgesparten Bereiche erfolgen. **Abb. 4b** zeigt eine alternative Anordnung der bestrahlten Flächen, die hier kreisförmig sind. Durch die Verwendung eines hexagonalen Hilfsgitters wird auch hier die Aussparung ausreichend großer Areale gewährleistet.

Abb. 5 zeigt das Prinzip der Oberflächenkühlung und die dadurch beeinflusste Modifizierung des Temperaturprofils. Die Oberflächenkühlung führt durch Wärmeentzug vor allem direkt unter der Oberfläche zu einer deutliche Abkühlung, so daß eine maximale Schonung der nicht zu schädigenden Strukturen möglich ist, so daß z. B. einer Narbenbildung vorgebeugt werden kann. Der prinzipielle Kurvenver-

lauf wird durch die Kühlung nicht beeinflusst, jedoch kann durch geeignete Anpassung der Bestrahlungsparameter (Laserleistung) eine im Vergleich zur ungekühlten Situation effizientere Erwärmung der interessierenden Bereiche erfolgen, die sich in einer erhöhten Temperatur im Bereich um 300 μ m Gewebetiefe widerspiegelt.

Abb. 6 zeigt das Absorptionsspektrum eines zur selektiven Erhöhung der Absorption geeigneten Farbstoffes, vorzugsweise ein für die Lebensmittelverarbeitung zugelassener Farbstoff (z. B. Indigo Carmin). Dieser besitzt ein ausgeprägtes Absorptionsmaximum bei 609 nm, so daß eine effiziente Ankopplung der Laserstrahlung im Bereich 600 nm gewährleistet ist. Dieser Wellenlängenbereich hatte sich als optimal zur Behandlung von subkutanen Gefäßanomalien herausgestellt, weil er einen guten Kompromiß zwischen Eindringtiefe und Ankoppeleffizienz an das Hämoglobin aufweist.

Abb. 7 zeigt das Absorptionsspektrum eines anderen Lebensmittelfarbstoffes mit einem Absorptionsmaximum bei 635 nm. Durch geeignete Wahl der Farbstoffe oder deren Mischung kann das Absorptionsmaximum der aktuellen Wellenlänge der Energiequelle angepaßt werden. Dies ist erforderlich, wenn z. B. eine Verlagerung des Intensitätsmaximums für bestimmte Hauttypen oder Hautveränderungen gemäß **Abb. 1** erfolgen soll.

Abb. 8 zeigt die prinzipielle Anordnung der Bestrahlungsgeometrie bei Benutzung einer Hochintensitätslichtquelle (1, z. B. Laser), einem optionalen Umlenkspiegel (2), einem zur Fokussierung oder Strahlaufweitung verwendeten optischen Element (3), mit der Anordnung einer geeigneten Leuchtfeldblende (4) zur erfindungsgemäßen Begrenzung des Bestrahlungsareals und dem zu bestrahlenden Gewebe (5).

Abb. 9 zeigt die Veränderung des Tiefenprofils der Intensitätsverteilung durch mechanische Kompression oder Dekompression der Hautoberfläche. Durch die mechanische Kraft können die optischen Parameter in kleinen Grenzen verändert werden, was sich direkt auf das Tiefenprofil der Intensitätsverteilung auswirkt. Erfindungsgemäß kann dieser Effekt zur Steuerung der Eindringtiefe genutzt werden.

Patentansprüche

1. Verfahren und Vorrichtungen zur Struktur bzw. tiefselaktiven Denaturierung biologischen Gewebes mittels Licht, **dadurch gekennzeichnet**, daß zur Einstellung der effektiven Wirtiefe mehrere technische Prozeßschritte wahlweise simultan, einzeln oder in freier Kombination durchgeführt werden. Dabei sind diese Prozeßschritte im einzelnen folgende:

- Druckaufschlagung des zu therapierenden Gewebearals.
- Einstellung der zur Therapie benutzten Belichtungsfläche.
- Einstellung der Wellenlänge des benutzten Lichtes.
- Kühlung der belichteten Gewebeoberfläche.
- Einstellung des Brechungsindexunterschiedes zwischen Gewebeoberfläche und Umgebung.

2. Verfahren und Vorrichtung nach 1, dadurch gekennzeichnet, daß die Druckbeeinflussung des Gewebes durch eine Gewebekompression mit voreinstellbarer Flächenbelastung erfolgt.

3. Verfahren und Vorrichtung nach 1, dadurch gekennzeichnet, daß die Druckbelastung des Gewebes durch die Anbringung eines Unterdruckes mittels Saugglocke erfolgt.

4. Verfahren und Vorrichtung nach 1–3, dadurch ge-

kennzeichnet, daß zur Einstellung der beleuchteten Wirkfläche eine Leuchtfeldblende im Abbildungsstrahlengang vorgesehen ist.

5. Verfahren und Vorrichtung nach 1–3, dadurch gekennzeichnet, daß zur Einstellung der beleuchteten Fläche ein optischer Wellenleiter, vorzugsweise ein für die Wirkwellenlänge transparenter Festkörper, verwendet wird.

6. Verfahren und Vorrichtung nach 1–3 und 5, dadurch gekennzeichnet, daß der optische Wellenleiter aus einem Faserbündel besteht.

7. Verfahren und Vorrichtung nach 1–3 und 5, dadurch gekennzeichnet, daß der optische Wellenleiter aus einem lichtleitenden Startmaterial besteht.

8. Verfahren und Vorrichtung nach 1–3 und 5, dadurch gekennzeichnet, daß der optische Wellenleiter ein Hohlleiter ist.

9. Verfahren und Vorrichtung nach 1–3, 5 und 8, dadurch gekennzeichnet, daß der Hohlleiter am distalen Ende durch ein Austrittsfenster abgeschlossen ist.

10. Verfahren und Vorrichtung nach 1–3, 5–9, dadurch gekennzeichnet, daß die Endfläche des Wellenleiters gleichzeitig zum Einbringen der Kompressionskraft benutzt wird.

11. Verfahren und Vorrichtung nach 1–10, dadurch gekennzeichnet, daß die Grenzschicht zwischen optischem Wellenleiter bzw. beleuchteter Fläche und Gewebeoberfläche mit einem Fluid einstellbarer Brechzahl hergestellt wird.

12. Verfahren und Vorrichtung nach 1–11, dadurch gekennzeichnet, daß das Fluid Wasser ist.

13. Verfahren und Vorrichtung nach 1–12, dadurch gekennzeichnet, daß das Fluid eine für die Wirkwellenlänge transparente Salbe oder Gel ist.

14. Verfahren und Vorrichtung nach 1–13, dadurch gekennzeichnet, daß die Kontaktfläche in voreinstellbarer Weise gekühlt wird.

15. Verfahren und Vorrichtung nach 1–14, dadurch gekennzeichnet, daß die Kühlung durch eine für die Wirkwellenlänge transparente Eisplatte erfolgt.

16. Verfahren und Vorrichtung nach 1, 3 und 14, dadurch gekennzeichnet, daß die Kühlung durch Aufsprühen eines Gas-Flüssig-Aerosols mit einer biokompatiblen niedrigsiedenden Flüssigkeit erfolgt.

17. Verfahren und Vorrichtung nach 1–16, dadurch gekennzeichnet, daß die Bestrahlung in einem regulären Muster in zwei Sitzungen durch Bestrahlung der jeweils übernächsten Nachbarfelder erfolgt.

18. Verfahren und Vorrichtung nach 1–17, dadurch gekennzeichnet, daß vor der Bestrahlung die zu behandelnde Gewebeoberfläche mit einem Lebensmittelfarbstoff behandelt wird, der sich durch Diffusion, Chemisorption und Kapillarkwirkung bevorzugt in Hautporen und Follikeln anreichert.

19. Verfahren und Vorrichtung nach 1–18, dadurch gekennzeichnet, daß vor der eigentlichen Behandlung die Hautoberfläche von Rückständen des Farbstoffes nach 18 gereinigt wird.

20. Verfahren und Vorrichtung nach 1–19, dadurch gekennzeichnet, daß das Absorptionsmaximum des Farbstoffes an die für eine vorgewählte Wirtiefe optimierte Wellenlänge angepaßt ist.

- Leerseite -

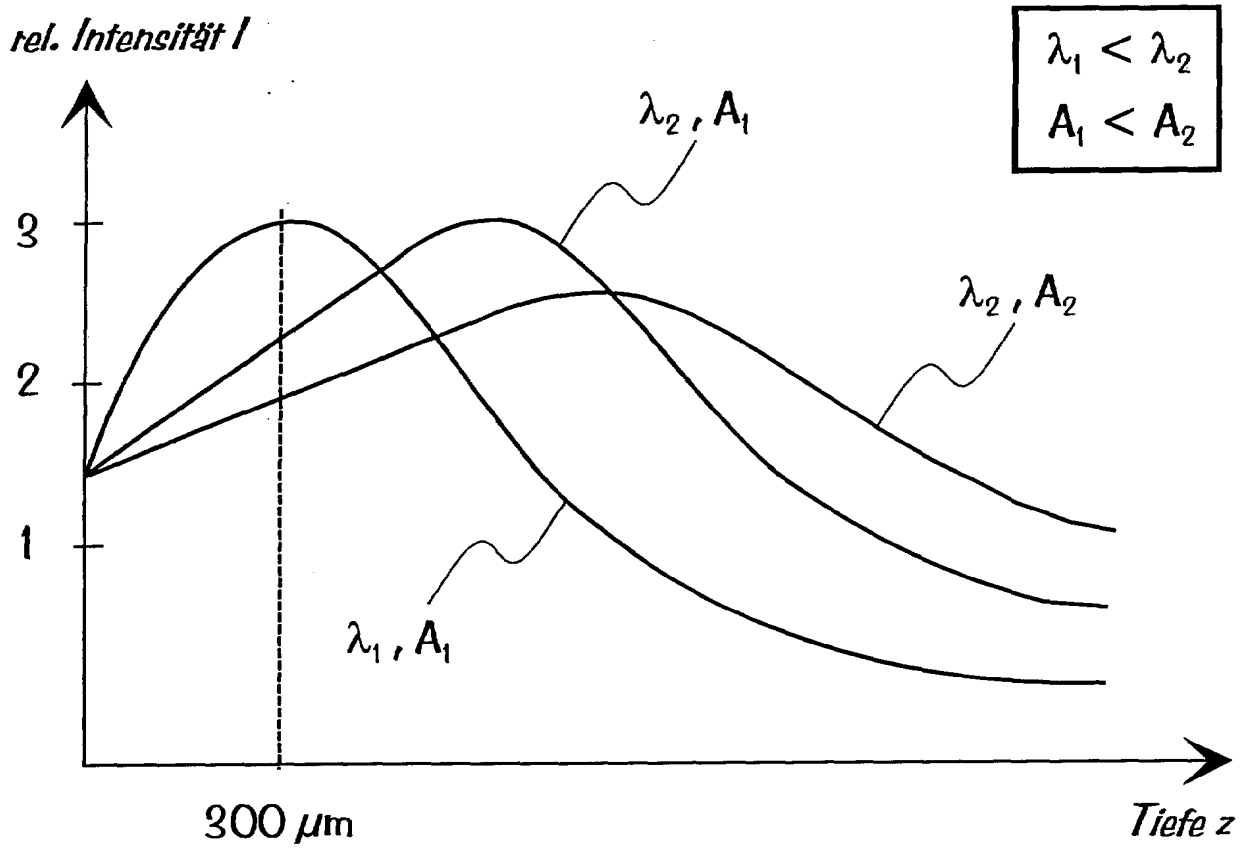
Fig. 1

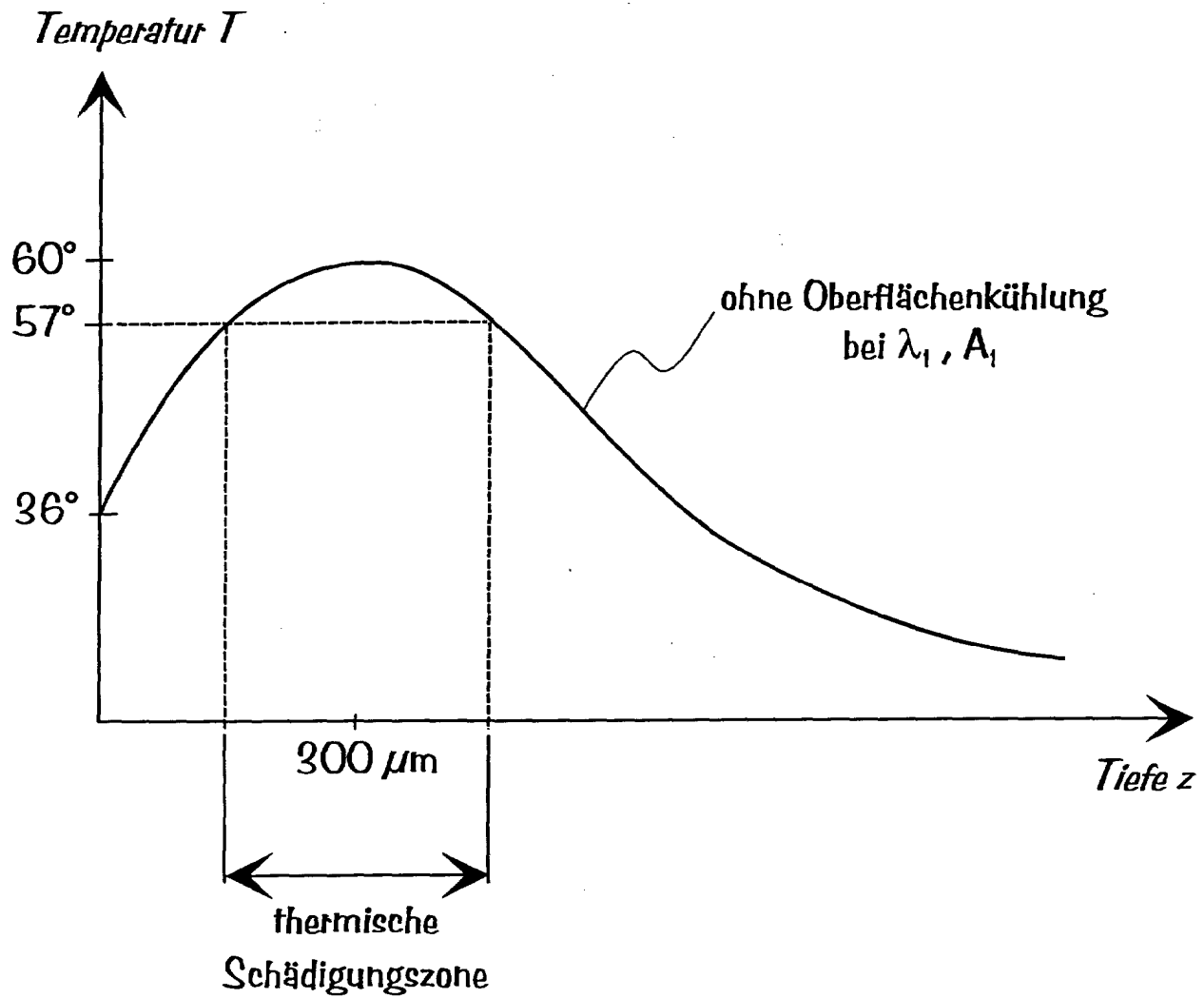
Fig. 2

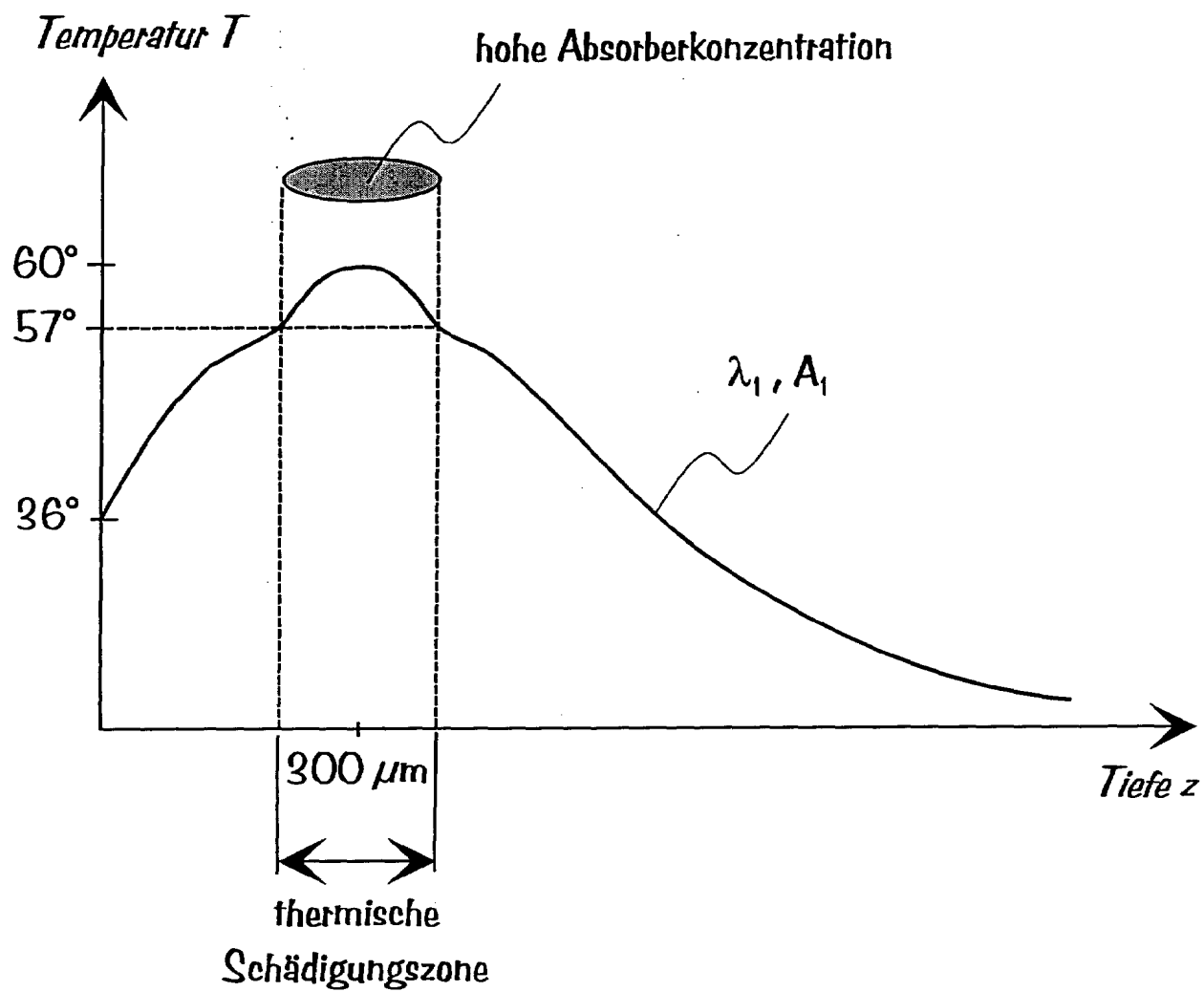
Fig. 3

Fig. 4a

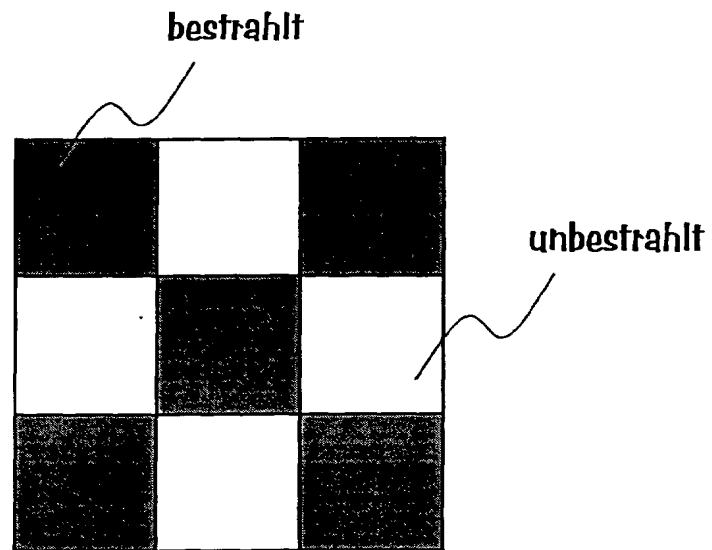


Fig. 4b

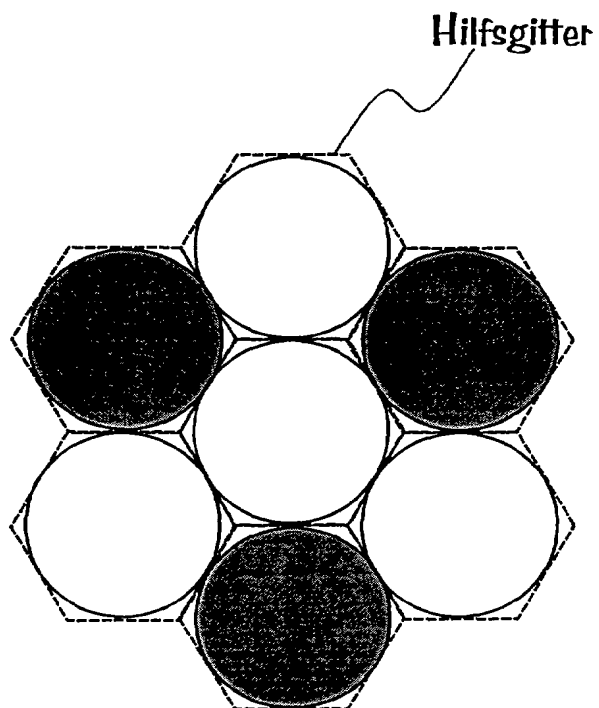


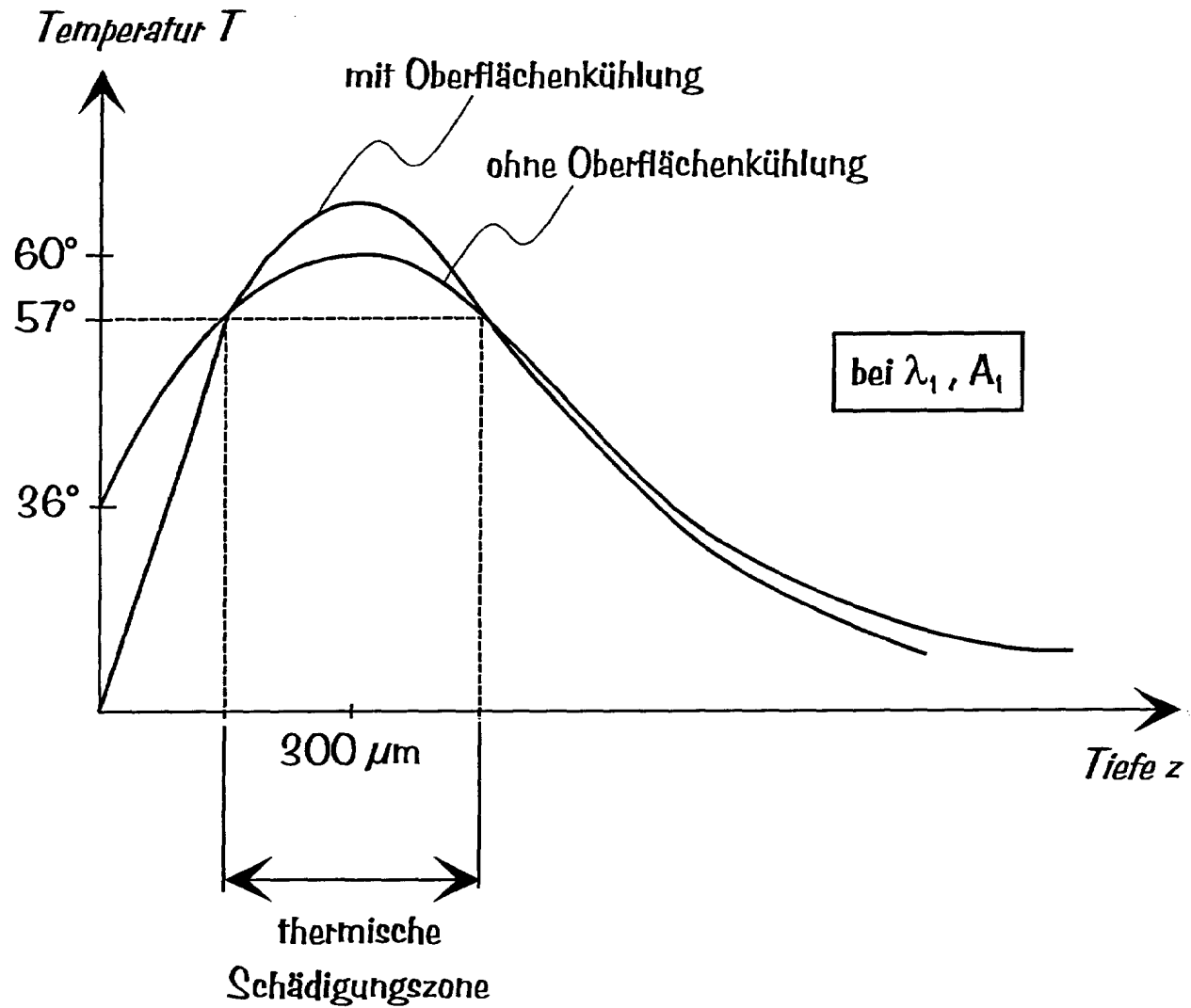
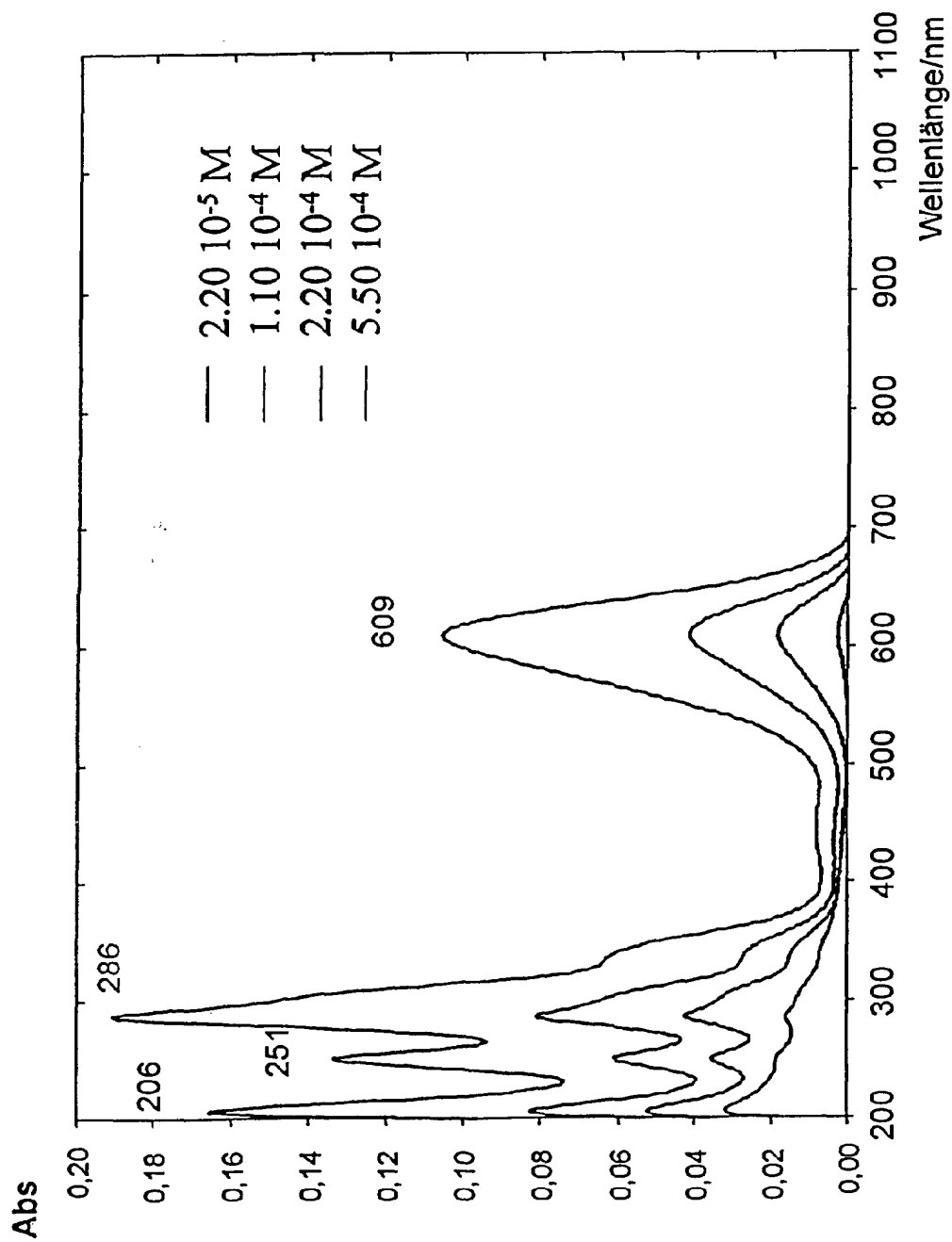
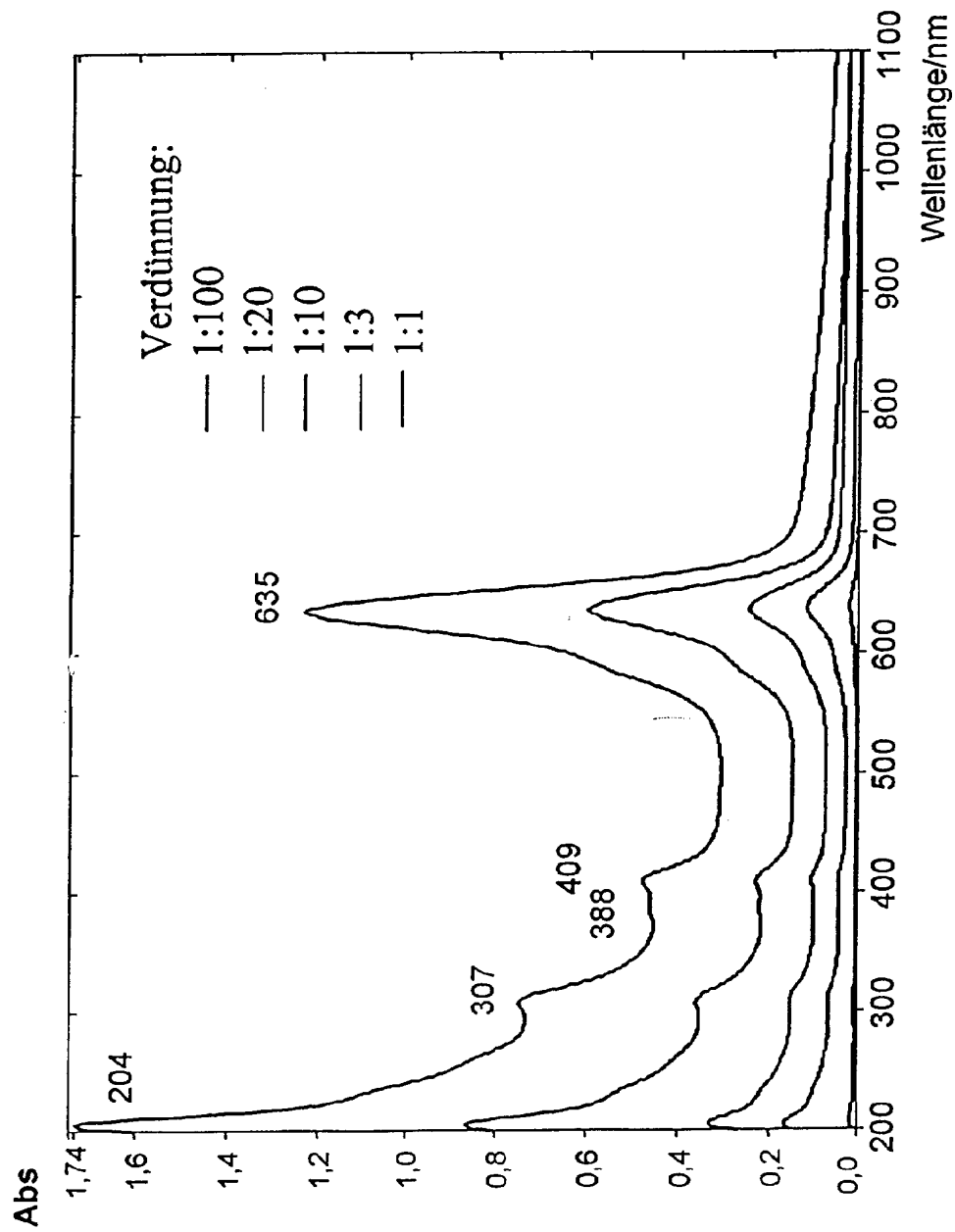
Fig. 5

Abb. 6



Indigo Carmin 25.6 mg/25ml, 1 mm Schichtdicke

Abb. 7



Lebensmittelfarbe, blau (Patentblau V E131, Cochenillerot A E124),
0.5g Probe/10ml, 1 mm Schichtdicke

Fig. 8

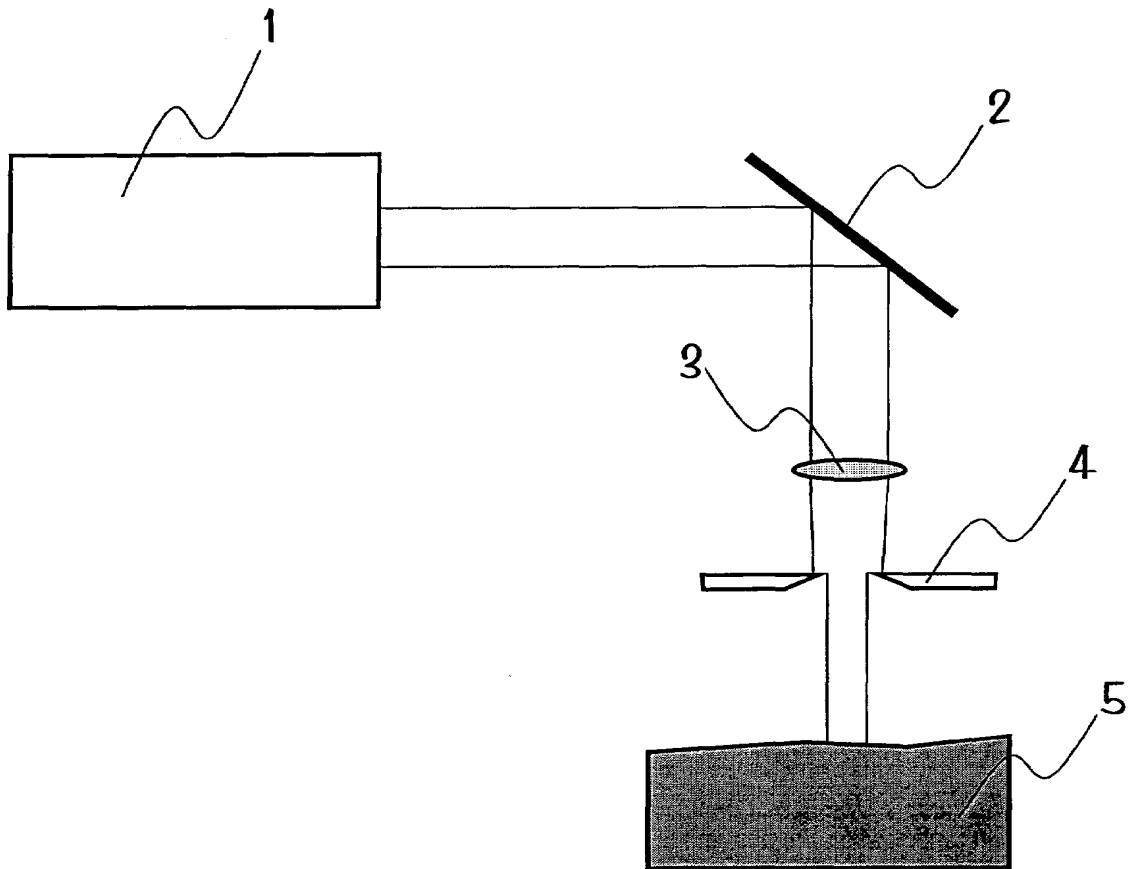
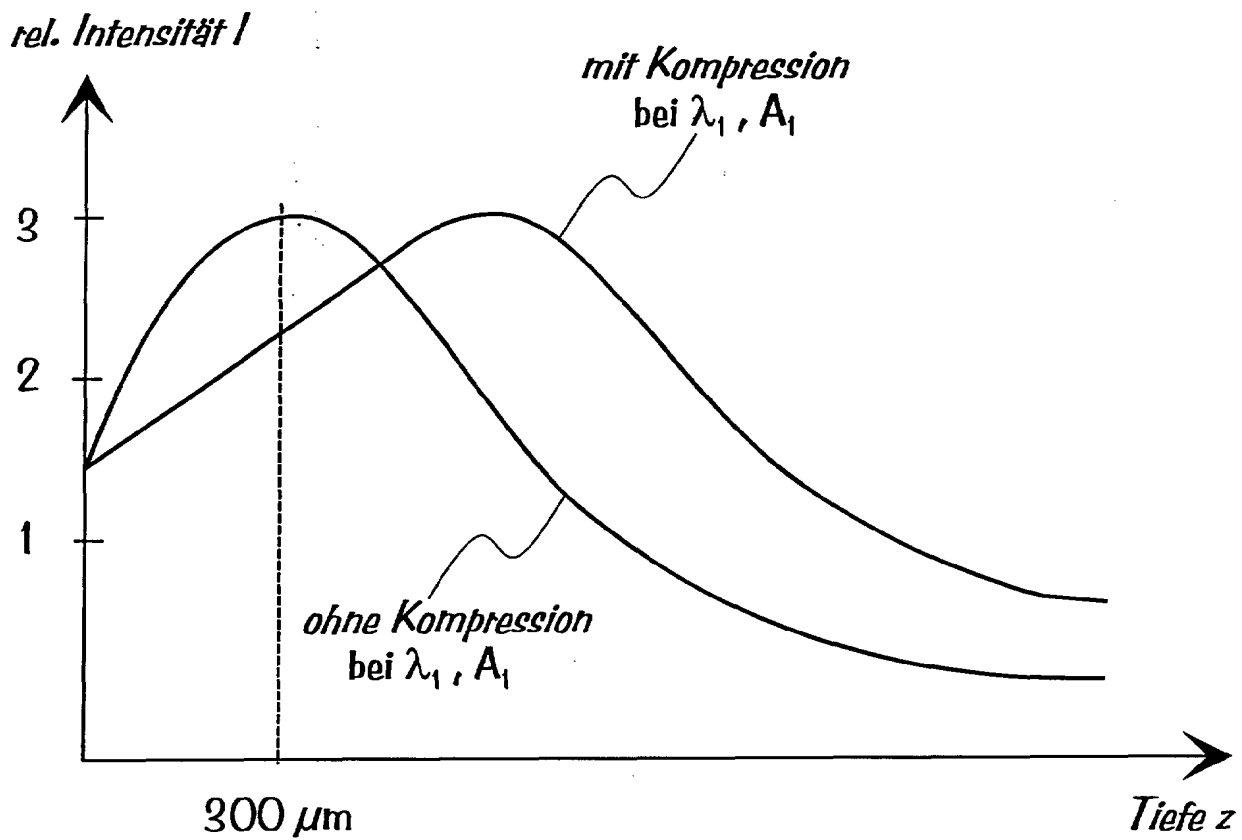



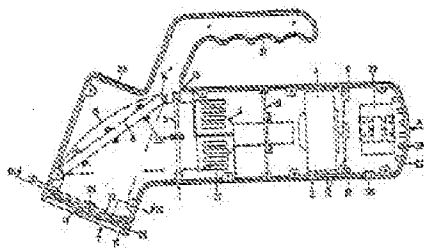


Fig. 9

AIR-COOLED APPARATUS FOR AN IRRADIATION BY MEANS OF POLARIZED LIGHT**Publication number:** DE3304230 (A1)**Publication date:** 1984-08-16**Inventor(s):** KLEIN FRANZ [DE]; BAUMANN OTTO [DE]**Applicant(s):** AMS AUTOMAT MESS & STEUERTECH [DE]**Classification:****- international:** **A61N5/06; F21V29/02; A61N5/00; A61N5/073; A61N5/06; F21V29/00; A61N5/00; (IPC1-7): A61N5/06****- European:** A61N5/06C2; F21V29/02**Application number:** DE19833304230 19830208**Priority number(s):** DE19833304230 19830208**Also published as:** WO8403049 (A1) EP0137005 (A1) EP0137005 (B1)**Abstract of DE 3304230 (A1)**

The unit designed particularly to stimulate the biological process related to cell activity, specially to enhance wound healing process at the surface of the body, comprises a casing (1, 2) wherein are arranged a light source (3), a fan (4) for the cooling, air inlet and outlet openings (18), as well as a light conducting medium (32) and filters (5) for the light and a medium (6) for polarizing the light. This unit should enable to reduce, respectively eliminate the adverse influences and/or perturbations of optical devices and/or damages caused to the person to be treated with the light by the air heated by the apparatus, respectively the cooling air so that at the end (7) of the casing turned towards the air inlet, an air permeable air filter, a filter support, a polarization apparatus, a light source and a fan may have an interchangeable arrangement, respectively an air permeable configuration; there are further provided guiding elements (15) for the air heated by the light source or blown by the fan and/or for the cooling air.



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**DEUTSCHES
PATENTAMT**

②1 Aktenzeichen: P 33 04 230.6
②2 Anmeldetag: 8. 2. 83
④3 Offenlegungstag: 16. 8. 84

DE 3304230 A1

⑦1 Anmelder:
ams Automatische Meß- und Steuerungstechnik
GmbH, 8572 Auerbach, DE

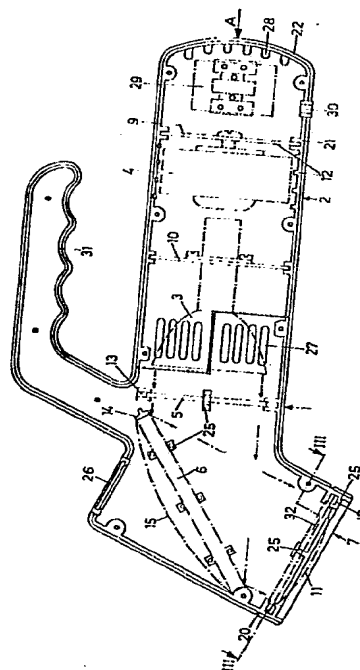
⑦2 Erfinder:
Klein, Franz, 8570 Pegnitz, DE; Baumann, Otto, 8572
Auerbach, DE

Beitrag zum

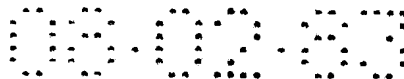
Prüfungsantrag gem. § 44 PatG ist gestellt

⑤4 **Bestrahlungsvorrichtung**

Die Vorrichtung, insbesondere zum Stimulieren biologischer Prozesse, die mit der Zellaktivität zusammenhängen, insbesondere zur Unterstützung des Heilprozesses von Verletzungen der Körperoberfläche, besteht aus einem Gehäuse, in welchem eine Lichtquelle, ein Ventilator zur Kühlung, Lufteintritts- und Luftaustrittsöffnungen sowie Lichtleitmittel und Filterelemente für das Licht und Mittel zum Polarisieren des Lichtes angeordnet sind. Mit dieser Vorrichtung sollen Beeinträchtigungen und/oder Störungen der optischen Einrichtungen und/oder Schädigungen des mit dem Licht zu behandelnden Individuums durch die vom Gerät erhitzte Luft bzw. durch die Kühlluft in der Weise vermindert bzw. ausgeschlossen werden, daß einmal an dem dem Lufteintritt zugewandten Ende des Gehäuses auswechselbar ein lichtdurchlässiges Luftfilterelement sowie Haltemittel für die Filterelemente, Polarisationsmittel, Lichtquelle und Ventilator luftdurchlässig angeordnet bzw. ausgebildet sind, wobei schließlich Führungsmittel für die durch die Lichtquelle erhitzte und vom Ventilator bewegte Luft und/oder für die Kühlluft vorgesehen sind.



DE 3304230 A1



3304230

PATENTANWALT DIPL.-PHYS. DR. K. SCHWEINZER

ZUGELASSENER VERTRETER VOR DEM EUROPÄISCHEN PATENTAMT

ESSENWEINSTRASSE 4-6 D-8500 NÜRNBERG 70 TELEFON 09 11 / 20 37 27 D TELEX 06 / 23 135

Nürnberg, 08.02.83
17/63A n s p r ü c h e :*Bestrahlungsv.*

1. Vorrichtung mit in einem Gehäuse (1,2) angeordneter Lichtquelle (3), einem Ventilator (4) zur Kühlung, Lufteintritts- und Luftaustrittsöffnungen, Lichtleitmittel (11), Filterelemente (5) für das Licht und Mittel
5 zum Polarisieren des Lichtes, insbesondere zur Stimulierung biologischer Prozesse, die mit der Zellaktivität zusammenhängen, insbesondere zur Unterstützung des Heilprozesses von Verletzungen der Körperoberfläche, wie Wunden, Ulcera und verschiedene epithele Schädigungen, aber auch für andere Anwendungen auf medi-
10 zinischem, pharmazeutischen und biologischem Gebiet, dadurch gekennzeichnet, daß am lufteintrittsseitigen Ende des Gehäuses (1,2) auswechselbar ein lichtdurchlässiges Luftfilterelement (8) angeordnet ist, daß die Haltemittel (10, 12, 13 und 14) für die Filterelemente
15 (5), Polarisationsmittel (6), Lichtquelle (3) und Ventilator (12) luftdurchlässig ausgebildet bzw. angeordnet sind und daß Führungsmittel (15) für die durch die Lichtquelle (3) erhitzte und vom Ventilator (4) bewegte
20 Luft und/oder die Kühlluft vorgesehen sind.

2. Vorrichtung nach Anspruch 1, dadurch gekennzeichnet, daß das Luftfilterelement (8) ein, eine zentrale Lichtdurchtrittsöffnung (16) umgebende, ringförmige
25

geändert gemäß Einlage
24.3.83

Luftöffnung (17) abdeckendes Filter (18) aufweist.

5 3. Vorrichtung nach Anspruch 1, dadurch gekennzeichnet,
daß das Luftfilterelement (9) ein sich über den Quer-
schnitt des Gehäuses (2) erstreckendes Filter aufweist.

10 4. Vorrichtung nach Anspruch 2 oder 3, dadurch gekenn-
zeichnet, daß das Filter (18) aus textilem Material
besteht.

5. Vorrichtung nach Anspruch 2 oder 3, dadurch gekenn-
zeichnet, daß das Filter (18) aus einem Drahtgewebe
besteht.

15 6. Vorrichtung nach Anspruch 2 oder 3, dadurch gekenn-
zeichnet, daß das Filter (18) aus Kunststoff-Fasern
besteht.

20 7. Vorrichtung nach Anspruch 2 oder einem der folgenden,
dadurch gekennzeichnet, daß das Luftfilterelement
(8) in einem auf das Gehäuse (1) aufsetzbaren Ring-
halter (19) angeordnet ist.

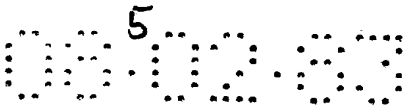
25 8. Vorrichtung nach Anspruch 7, dadurch gekennzeichnet,
daß das Gehäuse (1) eine Klemmnut für einen Ring-
halter (19) aufweist.

30 9. Vorrichtung nach einem der Ansprüche 3 bis 6,
dadurch gekennzeichnet, daß das Luftfilterelement
(9) in einem innerhalb des Gehäuses (2) vorgesehenen
Ringhalter (21) angeordnet ist.

35 10. Vorrichtung nach Anspruch 9, dadurch gekennzeich-
net, daß der Ringhalter (21) zwischen Ventilator (4)
und lufteintrittsseitigem Gehäuseende (22) angeordnet
ist.

11. Vorrichtung nach Anspruch 1 oder folgenden, dadurch gekennzeichnet, daß das Tragelement (10) für die Lichtquelle (3) aus einem Ringsockel (23) mit Luftschlitzen (24) besteht.
- 5
12. Vorrichtung nach Anspruch 1 oder folgenden, dadurch gekennzeichnet, daß das Tragelement (12) für den Ventilator (4) als Luftfilterelement ausgebildet ist.
- 10
13. Vorrichtung nach Anspruch 1 oder folgenden, dadurch gekennzeichnet, daß die Halterungen (13,14) für die optischen Elemente (5,6) mit schlitzartigen Luftdurchtrittsöffnungen versehen sind.
- 15
14. Vorrichtung nach einem oder mehreren der Ansprüche 1 bis 12, dadurch gekennzeichnet, daß die Haltemittel (13,14) für die optischen Elemente (5,6) und/oder die Tragelemente (10, 12) für die Lichtquelle (3) und/oder für den Ventilator (4) aus in Abständen am Umfang des Gehäuseinneren angeordneten Haltenocken (25)
- 20
- bestehen.
15. Vorrichtung nach einem oder mehreren der Ansprüche 1 bis 14, dadurch gekennzeichnet, daß im Lichtweg hinter dem Polarisationselement (6) ein Luftleitelement (15) vorgesehen ist.
- 25
16. Vorrichtung nach einem oder mehreren der Ansprüche 1 bis 15, dadurch gekennzeichnet, daß das Gehäuse (1) eine zur Lichtaustrittsrichtung parallele, luftdicht verschlossene Durchblicköffnung (26) aufweist.
- 30
17. Vorrichtung nach einem oder mehreren der Ansprüche 1 bis 16, dadurch gekennzeichnet, daß an beiden Endteilen (7, 22) des Gehäuses (1,2) je ein Luftfilterelement (8 bzw. 9) angeordnet ist.
- 35

18. Vorrichtung nach Anspruch 17, dadurch gekennzeichnet, daß der Ventilator (4) umschaltbar als Lüfter und Gebläse ausgebildet ist.



3304230

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Nürnberg, 08.02.83
17/63

Automatische Mess- und Steuerungstechnik GmbH,
Enge Gasse 1, 8572 Auerbach/Opf.

Bestrahlungsvorrichtung.

~~"Vorrichtung mit einer Lichtquelle, optischen-
Elementen, Mittel zum Polarisieren des Lichtes und
Ventilator, insbesondere für die Stimulierung
biologischer Prozesse."~~

geändert gemäß Eingabe
eingetragen: 29.3.83

Bestrahlungsv.

Die Erfindung betrifft eine Vorrichtung mit in einem Gehäuse angeordneter Lichtquelle, einem Ventilator zur Kühlung, Lufteintritts- und Luftaustrittsöffnungen, Lichtleitmittel, Filterelemente für das Licht und Mittel
5 zum Polarisieren des Lichtes, insbesondere zur Stimulierung biologischer Prozesse, die mit der Zellaktivität zusammenhängen, insbesondere zur Unterstützung des Heilprozesses von Verletzungen der Körperoberfläche, wie Wunden, Ulcera und verschiedene Epithel-
10 Schädigungen, aber auch für andere Anwendungen auf medizinischem, pharmazeutischen und biologischem Gebiet.

Es ist bereits ein Verfahren zur Stimulierung biologischer Prozesse bekannt, bei dem der zu stimulierende Bereich
15 mit einem Lichtbündel vorbestimmter Intensität bestrahlt wird, und die Bestrahlung mit einem linear polarisierten nicht kohärenten Licht erfolgt, das Wellenlängen-Komponenten oberhalb von 300 nm aufweist. Vorzugsweise hat das bestrahlende Licht eine kontinuierliche oder
20 quasi kontinuierliche spektrale Verteilung wenigstens im Wellenlängenbereich von größer 300 nm. Bei einer Vorrichtung zur Erzeugung des benötigten Lichtes ist beispielsweise eine Lichtquelle vorgesehen, die inkohärentes Licht mit Spektralkomponenten oberhalb von 300 nm
25 abgibt, und ein Ablenssystem im Strahlengang des von der Lampe abgegebenen Lichtes um die Lichtstrahlen in eine gegebene Behandlungsrichtung zu bringen und im Strahlengang ein Polarisator um polarisierte Lichtstrahlen zu erzeugen.

30 Die Lichtquelle erzeugt, physikalisch bedingt, Wärme, die aus dem die Lichtquelle aufnehmenden Gehäuse abgeführt werden muß. Dazu ist ein Ventilator vorgesehen, der die Lichtquelle kühlt. Durch den Ventilator
35 wird Luft in das Gehäuse eingesaugt bzw. aus dem

Gehäuse ausgeblasen. Dadurch besteht die Gefahr einer Verschmutzung der optischen Einrichtungen durch Staub oder einer Kontamination durch mit der Luft mitgerissene Bakterien, Viren oder dergl.

5

Der Erfindung liegt die Aufgabe zugrunde, eine Vorrichtung der eingangs erläuterten Art dahingehend auszubilden, daß Beeinträchtigungen und/oder Störungen der optischen Einrichtungen und/oder Schädigungen des mit dem Licht zu behandelnden Individuums durch die vom Gerät erzeugte warme Luft bzw. die Kühlluft vermindert bzw. ausgeschlossen werden.

10

Diese Aufgabe wird nach der Erfindung bei der Vorrichtung der eingangs erläuterten Art im wesentlichen dadurch gelöst, daß am lufteintrittsseitigem Ende des Gehäuses auswechselbar ein lichtdurchlassiges Luftfilterelement angeordnet ist, daß die Haltemittel für die Filterelemente, Polarisationsmittel, Lichtquelle und Ventilator luftdurchlässig ausgebildet bzw. angeordnet sind und daß Führungsmittel für die durch die Lichtquelle erhitzte und vom Ventilator bewegte Luft und/oder die Kühlluft vorgesehen sind.

15

20

25

30

Gemäß einem ersten Ausführungsbeispiel der Erfindung weist das Luftfilterelement ein, eine zentrale Lichtdurchtrittsöffnung umgebende, ringförmige Luftöffnung abdeckendes Filter auf. Es ist sichergestellt, daß das Filter weder den Lichtstrahl stört noch durch den Lichtstrahl beaufschlagt wird. Die vom Ventilator angesaugte Kühlluft streicht dann im wesentlichen an der Innenwand des Gehäuses entlang und verläßt das Gehäuse am rückwärtigen Ende.

35

In manchen therapeutischen Fällen oder auch anderen Anwendungen kann es vorteilhaft sein, die zu behan-

delnde Fläche nicht nur mit Licht zu bestrahlen, sondern auch zu erwärmen. Dazu kann in vorteilhafter Weise die von der Lichtquelle erwärmte Luft verwendet werden, die dann vom Ventilator durch die vordere, luftaustritts-

5 seitige, Öffnung des Gehäuses abgeblasen wird. In diesem Fall ist es vorteilhaft, wenn das Luftfilterelement bezüglich des Strahlenganges des Lichtes hinter der Lichtquelle angeordnet ist. In vorteilhafter Weise weist das Luftfilterelement ein sich über den Querschnitt des

10 Gehäuses erstreckendes Filter auf.

Das jeweilige Luftfilter kann in verschiedener Weise realisiert sein. Bei einem ersten Ausführungsbeispiel besteht das Filter aus textilem Material. Bei einer

15 Abwandlung der Erfindung besteht das Filter aus einem Drahtgewebe. In noch weiterer Abwandlung kann das Filter aus Kunststoff-Fasern oder dergl. bestehen.

Bei einem besonders vorteilhaften Ausführungsbeispiel ist das Luftfilterelement in einem auf das Gehäuse aufsetzbaren Ringhalter angeordnet.

20

Zur Vereinfachung der Verbindung zwischen Ringhalter und Gehäuse weist das Gehäuse eine Klemmnut für

25 den Ringhalter auf.

Bei dem Ausführungsbeispiel, bei dem der Ventilator die von der Lichtquelle erwärmte Luft vorne, d.h. parallel mit dem Lichtstrahl ausbläst, ist das Luftfilterelement

30 in einem innerhalb des Gehäuses vorgesehenen Ringhalter angeordnet.

Um den Ventilator in den Schutz gegen Verstaubung mit einzuschließen ist es vorteilhaft, wenn der Ringhalter zwischen Ventilator und lufteintrittsseitigem Gehäuse-

35 ende angeordnet ist.

In zweckmäßiger Ausgestaltung der Vorrichtung besteht das Tragelement für die Lichtquelle aus einem Ringsockel mit Luftschlitzen. Dadurch wird erreicht, daß die Lichtquelle vollständig von der Kühlluft umströmt wird.

5

Gemäß einer besonderen konstruktiven Ausgestaltung der erfindungsgemäßen Vorrichtung ist das Tragelement für den Ventilator zugleich als Luftfilterelement ausgebildet.

10

In zweckmäßiger Weiterbildung sind die Halterungen für die optischen Elemente mit schlitzartigen Luftdurchtrittsöffnungen versehen.

15

Gemäß einem bevorzugten Ausführungsbeispiel der Erfindung bestehen die Haltemittel für die optischen Elemente und/oder die Tragelemente für die Lampe und/oder dem Ventilator aus in Abständen am Umfang des Gehäuseinneren angeordneten Haltenocken.

20

Insbesondere als Polarisationsmittel Spiegel oder dergl. verwendet werden, ist es vorteilhaft, wenn im Lichtweg hinter dem Polarisationselement ein Luftleitelement vorgesehen ist.

25

In zweckmäßiger Weiterbildung der Erfindung weist das Gehäuse eine zur Lichtaustrittsrichtung parallele, luftdicht verschlossene Durchblicköffnung auf.

30

Um sowohl eine Betriebsart durchführen zu können, bei der die Kühlluft am lichtaustrittsseitigem Ende des Gehäuses angesaugt wird als auch eine Betriebsart, bei der aus dem lichtaustrittsseitigem Ende Warmluft ausgeblasen wird, ist es vorteilhaft, wenn an beiden

35

Endteilen des Gehäuses je ein Luftfilterelement angeordnet ist.

5 Für dieses Ausführungsbeispiel ist es ferner vorteilhaft, wenn der Ventilator umschaltbar als Lüfter und Gebläse ausgebildet ist.

10 Weitere Einzelheiten, Merkmale und Vorteile der Erfindung werden anhand der Zeichnung näher erläutert, die schematisch ein Ausführungsbeispiel darstellt.

Dabei zeigt:

15 Fig. 1 einen Querschnitt durch ein Ausführungsbeispiel einer erfindungsgemäßen Vorrichtung,

Fig. 2 eine Rückansicht in Richtung des Pfeiles A der Fig. 1,

20 Fig. 3 eine ~~Stirn~~^Hansicht in Richtung des ~~Pfeiles B~~^{Schnittlinie III-III} der Fig. 1,

Fig. 4 einen Teilausschnitt des Gehäuses mit aufgestecktem Luftfilterelement, und

Fig. 5 ein Ausführungsbeispiel eines Tragelementes.

25 In einem, beispielsweise zweiteilig ausgebildeten, Gehäuse mit einem vorderen Gehäuseteil 1 und einem rückwärtigem Gehäuseteil 2 ist eine Lichtquelle 3 angeordnet, die das für die beispielsweise Stimulierung biologischer Prozesse bestimmte Licht abgibt. Ferner sind
30 Lichtfilter, beispielsweise UV-und/oder IR-Filter 5, ein Polarisator, beispielsweise ein Brewster-Spiegel 6 sowie ein Ventilator 4 angeordnet. Ferner können Lichtleitmittel, Spiegel oder Linsen 32 vorgesehen sein. Im Gehäuse ist ferner eine elektrische Anschlußplatte 29
35 vorgesehen, die durch eine Kabelzuführungsöffnung 30 ein entsprechendes Kabel für die Netzversorgung angeschlossen ist.

Am rückseitigen Ende des Gehäuses 2 sind Kühlschlitze 28 angebracht. Die Gehäuseteile 1, 2 können vorzugsweise im Bereich der Lichtquelle 3 Kühlschlitze 27 aufweisen.

5 Das Gehäuse 2 bzw. Gehäuseteil 1 kann einen Handgriff 31 aufweisen bzw. mit einem Handgriff 31 einstückig ausgebildet sein. Anstelle des Handgriff 31 kann auch eine Haltevorrichtung für eine Stativbefestigung oder dergl. angebracht bzw. vorgesehen sein. Am lichtaustritts-
10 seitigem Ende 7 des Gehäuses 1 ist eine Lichtleit bzw. Sammelvorrichtung, beispielsweise eine Linse 32 vorgesehen.

Zur Beobachtung des Behandlungsfeldes ist zweckmäßiger
15 Weise am Gehäuseteil 1 eine Durchblicksöffnung 26 vorgesehen, deren Blickrichtung parallel zum Lichtstrahl verläuft. Die Durchblicksöffnung 26 ist luftdicht verschlossen.

20 Um das Eindringen von Staub und/oder Bakterien oder dergl. zu verhindern, ist erfindungsgemäß am vorderen lichtaustrittsseitigem Ende 7 des Gehäuses 1 ein Luftfilterelement 8 angebracht. Dieses Luftfilterelement 8 besteht vorzugsweise, gemäß Fig. 3, aus einem Filter 18
25 der ringförmige Luftaustrittsöffnungen 17 überdeckt. Der ringförmige Teil 17 umgibt eine zentrale Lichtdurchtrittsöffnung 16. Die vom Ventilator 4 angesaugte Kühlluft strömt durch das, die ringförmige Luftdurchtrittsöffnung 17 abdeckende, Filter 18 und wird dabei ent-
30 staubt. Das Filter 18 kann aus textilem Material bestehen, beispielsweise einem feingewebtem Tuch, oder aus einem Metallgitter mit sehr feinen Gitteröffnungen, oder aus einem Kunststofffilter in Tuchform aus Fasern hergestellt oder aus feinem Granulat oder dergl. Bei
35 der Verwendung von Textilgewebe besteht die Möglichkeit, dieses bei Gebrauch mit einer desinfizierenden Lösung

zu beaufschlagen. Dabei wird sichergestellt, daß Viren und Bakterien oder dergl. nicht in das Gehäuse-Innere eindringen und diese kontaminieren können.

5 Durch die angesaugte Luft wird die Lichtquelle 3 gekühlt, wobei die erwärmte Luft durch die Kühlöffnungen 28 am hinteren Ende des Gehäuses 2 ausgeblasen wird.

10 Um ein leichtes Auswechseln des Luftfilterelementes zu ermöglichen, ist es vorteilhaft, wenn das Element 8 in einem Ringhalter 19 angeordnet ist, der über das vordere Ende 7 des Gehäuses 1 geschoben werden kann. Zur sicheren Verbindung mit dem Gehäuse 1 ist es zweckmäßig, wenn das Gehäuse eine Klemmnut 20 auf-
15 weist (Fig. 4).

Ein Haltemittel 10 für die Lichtquelle 3 ist (Fig. 5) derart ausgebildet, daß ein Ringsockel 23 von Luftschlitzen 24 umgeben ist. Dadurch wird erreicht, daß
20 die vom Ventilator 4 angesaugte Kühlluft bzw. die vom Ventilator 4 ausgeblasene Kühlluft die Lichtquelle 3 vollständig umfließen kann.

25 Die Halteelemente 13 für das Filterelement 5, 14 für den Polaristor 6 sind analog zum Luftfilterelement 8 mit einer zentralen Lichtdurchtrittsöffnung und ringförmigen Luftdurchtrittsöffnungen versehen.

30 Die Halteelemente können in ringförmigen Halterungen 21 befestigt sein, oder in einzelnen, über den Umfang des Gehäuses verteilten Haltenocken 25.

35 Die Haltemittel 12 für den Ventilator 4 können gleichzeitig, wie in Fig. 1 dargestellt, als Luftfilterelement 9 ausgebildet sein.

Bei einer ersten Betriebsart der erfindungsgemäßen
 Vorrichtung wird vom Ventilator 4 am lichtaustrittsseitigem
 Ende 7 des Gehäuses 1 Luft durch ein vorderes Luft-
 filterelement 8 angesaugt und durch das Gehäuse,
 5 an der Lichtquelle 3 vorbei, durchgeleitet und am
 Gehäuseende durch die Kühlschlitze 28 abgeführt.

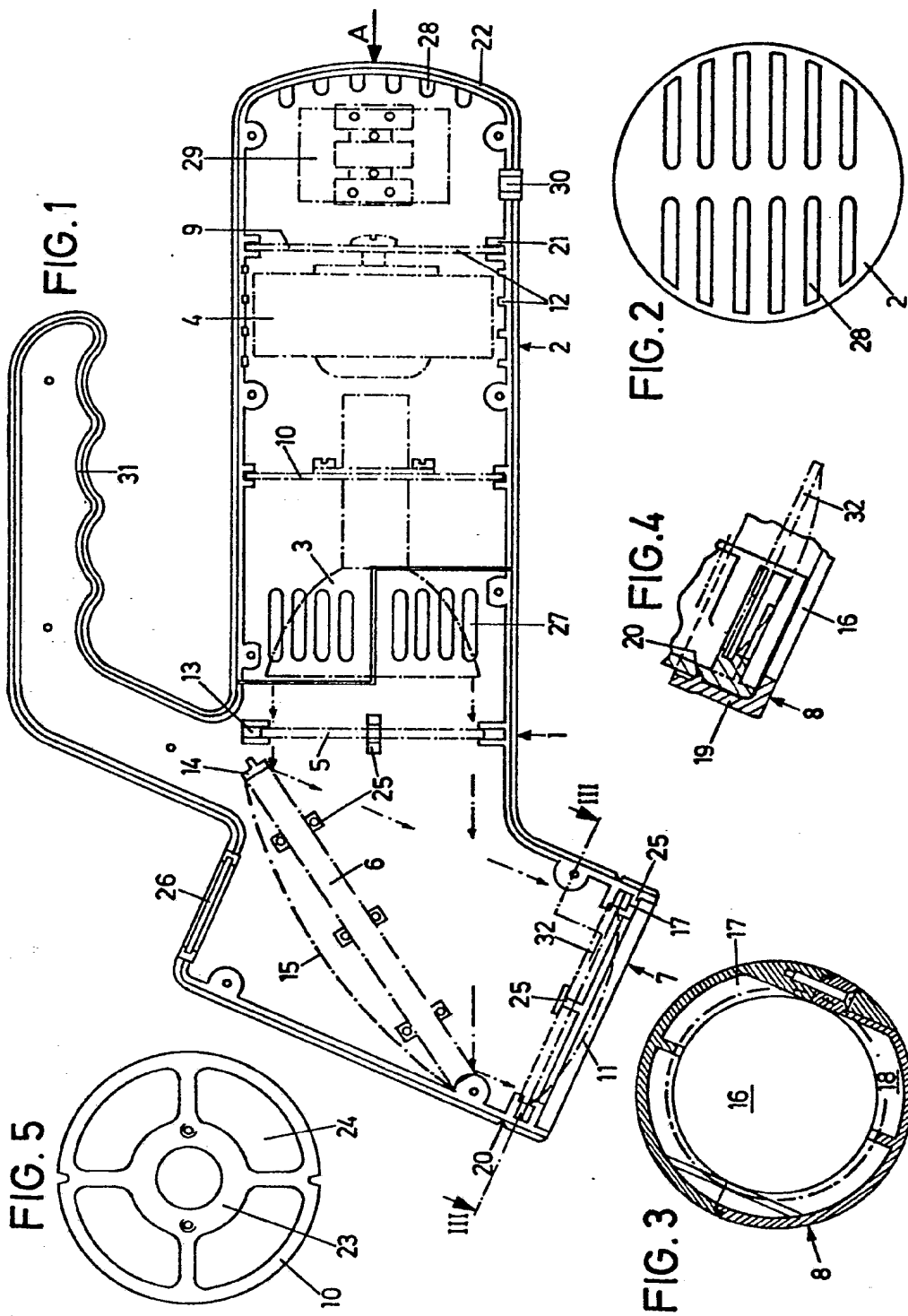
Bei einer zweiten Betriebsart saugt der Ventilator
 4 Luft durch die Öffnungen 28 am rückwärtigen Teil an,
 10 bläst sie über die Lichtquelle 3 wobei die Luft erhitzt
 wird und drückt die Luft dann durch das luftaustritts-
 seitige Ende 7, parallel zum Lichtstrahl, auf die zu
 behandelnde Fläche. Bei dieser zweiten Betriebsart
 wird der therapeutische Effekt der Lichtbestrahlung
 15 durch die lokale Erwärmung mittels der von der Licht-
 quelle 3 erhitzten Luft unterstützt.

Die Erfindung ist nicht auf das dargestellte und be-
 schriebene Ausführungsbeispiel beschränkt. Bei Verwen-
 20 dung eines vom Licht geradlinig durchlaufenen Polari-
 sator 6 kann das Gehäuse als geradliniges Rohr ausge-
 bildet sein. Das Gehäuse kann einteilig sein. Die Kühl-
 schlitze 27 können entfallen. Das Gehäuse kann kreisförm-
 25 igen oder auch rechteckigen, quadratischen oder ovalen
 Querschnitt aufweisen. Die Luftführungsmittel 15 können,
 wenn sie aus durchsichtigem Material bestehen, auch
 vor dem Polarisator 6 angeordnet sein. Die Erfindung
 umfaßt somit alle fachmännischen Weiterbildungen und
 Abwandlungen sowie Teil- und/oder Unterkombinationen
 30 der beschriebenen und/oder dargestellten Merkmale
 und Maßnahmen.

1	Gehäuse	42
2	Gehäuse	43
3	Lichtquelle	44
4	Ventilator	45
5	UV-/IR-Filter	46
6	Polarisator	47
7	lichtaustrittsseitiges Ende von 1	48
8	Luftfilterelement (vorne)	49
9	Luftfilterelement (hinten)	50
0	Haltemittel für Lichtquelle	51
1	Lichtleitelement	52
2	Haltemittel für Ventilator	53
3	Haltemittel für Filterelement	54
4	Haltemittel für Polarisator	55
5	Luftführungsmittel	56
6	zentrale Lichtdurchtrittsöffnung	57
7	ringförmige Luftdurchtrittsöffnung	58
8	Filter	59
9	Ringhalter	60
20	Klemmnut	61
21	Ringhalter	62
22	lufteintrittsseitiges Gehäuseende	63
23	Ringsockel	64
24	Luftschlitz	65
25	Haltenocken	66
26	Durchblicksöffnung	67
27	Kühlschlitz	68
28	Kühlschlitz	69
29	elektrische Anschlußplatte	70
30	Kabelzuführungsöffnung	71
31	Handgriff	72
32	Linse	73
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34		75
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36		77
37		78
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40		81
41		82

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Medical laser handpiece

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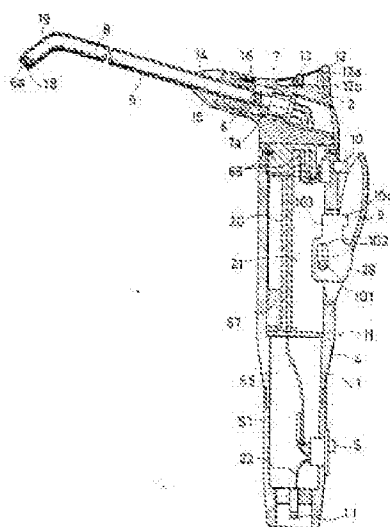
US4617926 (A)

US4619612 (A)

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Abstract of **US 4826431 (A)**

A medical laser handpiece comprising a grip body, a semiconductor laser generator disposed in the grip body and an irradiation nozzle which is detachably cross-connected to the head of the grip body at an angle and includes a laser light transmitting means from the semiconductor laser generator. With this laser handpiece, an operator can accurately and easily treat even relatively small and complicated shaped teeth and periodontal sections in the mouth. In addition to a structure capable of irradiating laser light and a structure capable of cooling the semiconductor generator, this invention also includes a structure capable of jetting air, water and a mist of air and water, and a structure capable of reducing irradiation loss of laser light.



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PATENTAMT

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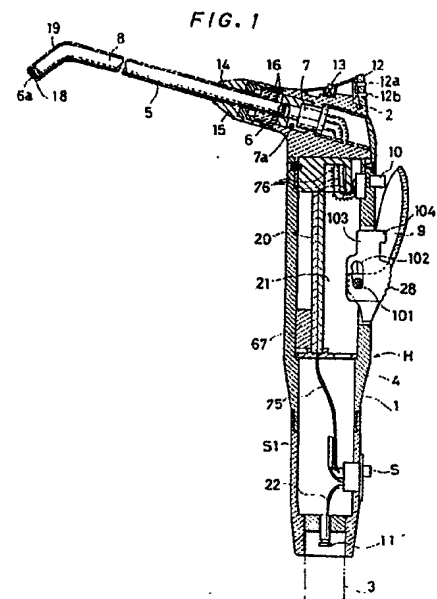
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⑤④ Medizinisches Laserhandstück

Medizinisches Laserhandstück mit einem Griffkörper, einem innerhalb des Griffkörpers angeordneten Halbleiter-Lasergenerator und einer mit dem Kopf des Griffkörpers lösbar verbundenen, gegenüber dem Griffkörper abgewinkelten Bestrahlungsdüse, die mit einer Übertragungseinrichtung zum Übertragen von von dem Halbleiter-Lasergenerator kommendem Laserlicht versehen ist. Vorzugsweise sind ferner Mittel zum Kühlen des Halbleiter-Lasergenerators, zum Abgeben eines Strahls aus Luft, Wasser oder einem Luft/Wasser-Nebel sowie Mittel zum Verringern von Laserlichtverlusten vorgesehen.



DE 37 19561 A 1

Patentansprüche

1. Medizinisches Laserhandstück, **gekennzeichnet durch** einen Griffkörper (1), einen innerhalb des Griffkörpers angeordneten Halbleiter-Lasergenerator (7) und eine mit dem Kopf (2) des Griffkörpers lösbar verbundene, gegenüber dem Griffkörper abgewinkelte Bestrahlungsdüse (5), die mit einer Übertragungseinrichtung (8) zum Übertragen von von dem Halbleiter-Lasergenerator kommenden Laserlicht versehen ist.
2. Medizinisches Laserhandstück nach Anspruch 1, dadurch gekennzeichnet, daß der Halbleiter-Lasergenerator (7) auf der Achse der Bestrahlungsdüse (5) in dem Griffkopf (2) sitzt.
3. Medizinisches Laserhandstück nach Anspruch 1 oder 2, dadurch gekennzeichnet, daß der Griffkopf (2) mit einer Beleuchtungsvorrichtung (23) versehen ist und die Laserlicht-Übertragungseinrichtung (8) sowohl dem Übertragen von Beleuchtungslicht von der in dem Griffkörper (1) untergebrachten Beleuchtungsvorrichtung als auch dem Übertragen von Laserlicht von dem in dem Griffkopf angeordneten Halbleiter-Lasergenerator (7) dient.
4. Medizinisches Laserhandstück nach einem der vorhergehenden Ansprüche, dadurch gekennzeichnet, daß auf dem Griffkopf (2) eine Anzeigeeinrichtung (12) zur Anzeige des Laserbestrahlungs-Bereitschaftszustands, des Laserbestrahlungszustands und der Laserbestrahlungs-Zeitwahl vorgesehen ist.
5. Medizinisches Laserhandstück nach einem der vorhergehenden Ansprüche, dadurch gekennzeichnet, daß in den Griffkörper (1), den Griffkopf (2) und die Bestrahlungsdüse (5) Luft- und Wasserzuleitungen (24, 25, 26, 27) eingebaut sind.
6. Medizinisches Laserhandstück nach einem der vorhergehenden Ansprüche, dadurch gekennzeichnet, daß die Bestrahlungsdüse (5) an ihrem Ende mit einem abgebogenen Abschnitt (19) versehen und mit dem Griffkopf (2) drehbar verbunden ist.
7. Medizinisches Laserhandstück nach einem der vorhergehenden Ansprüche, dadurch gekennzeichnet, daß der Halbleiter-Lasergenerator (7) in dem Griffkopf (2) über ein mit Kühlrippen (30) versehenes Halteteil (29) befestigt ist, das eine Luftkühlung des Halbleiter-Lasergenerators und des diesen umgebenden Bereichs erlaubt.
8. Medizinisches Laserhandstück nach Anspruch 7, dadurch gekennzeichnet, daß eine Luftzuleitung (24) in den Griffkörper (1) eingebaut und ein mit dieser Luftzuleitung verbundener Zwischenraum (37) zwischen dem Griffkörper und dem Halbleiter-Lasergenerator (7) ausgebildet ist.
9. Medizinisches Laserhandstück nach Anspruch 7 oder 8, dadurch gekennzeichnet, daß das Halteteil (29) aus einem Werkstoff mit hoher Wärmeleitfähigkeit gefertigt ist.
10. Medizinisches Laserhandstück nach einem der vorhergehenden Ansprüche, dadurch gekennzeichnet, daß die Laserlicht-Übertragungseinrichtung (8) in ein Schutzrohr (81) eingesetzt ist, daß ein wasserführender Zwischenraum (56) zwischen der Laserlicht-Übertragungseinrichtung und dem Schutzrohr vorgesehen ist, daß luftführende Zwischenräume (53, 55) zwischen dem Schutzrohr und dem Griffkopf (2) sowie zwischen dem Schutzrohr und der Bestrahlungsdüse (5) vorgesehen sind, und daß

diese Zwischenräume derart ausgebildet sind, daß Wasser und Luft nach Durchlaufen der Zwischenräume aus dem Ende der Bestrahlungsdüse als Strahl austreten können.

11. Medizinisches Laserhandstück nach einem der vorhergehenden Ansprüche, dadurch gekennzeichnet, daß zwischen dem Halbleiter-Lasergenerator (7) und der Laserlicht-Übertragungseinrichtung (8) ein Laserlicht-Aufnahmeabschnitt (39) mit einer sich nach vorne verjüngenden, totalreflektierenden Innenfläche angeordnet ist.

Beschreibung

Die Erfindung betrifft ein medizinisches Handstück, das mit einem Laserlicht erzeugenden Halbleiter-Lasergenerator ausgerüstet ist, um Laserlicht mit hoher Genauigkeit auf zu behandelnde Stellen auftreffen zu lassen. Bei der Behandlung kann es sich unter anderem um eine entzündungshemmende Behandlung, um eine Schmerzlinderung, um eine Beschleunigung eines Heilungsprozesses oder um das Aushärten von optisch polymerisierbaren Substanzen handeln.

Es sind Laserhandstücke bekannt, die einen relativ großen Lasergenerator, beispielsweise einen YAG-Lasergenerator oder einen CO₂-Lasergenerator erfordern. Daneben sind ferner medizinische Laserhandstücke mit einem kleinen Halbleiter-Lasergenerator der vorliegend vorgesehenen Art bekannt. Im Falle der zuerst genannten Handstücke wird das von dem Lasergenerator erzeugte Laserlicht dem Handstück über eine Laserlicht-Übertragungseinrichtung, eine Spiegelanordnung oder einen Manipulator zugeführt. Bei den letztgenannten Handstücken ist ein Halbleiter-Lasergenerator in den Handstückkörper oder -griff eingebaut (JP-OS 60-24 832).

Wenn ein Laserhandstück zur Behandlung von relativ kleinen und kompliziert geformten Zahn- und Periodontalbereichen im Mund benutzt wird, muß das Ende oder der Kopf des Handstückkörpers, während dieser gehalten wird, gedreht werden, so daß das Laserlicht genau auf die vorgesehenen Zahn- und Periodontalbereiche auffällt. Das bekannte medizinische Laserhandstück mit im Griff des Handstückes untergebrachtem Halbleiter-Lasergenerator ist so aufgebaut, daß Laserlicht auf relativ breite und flache Bereiche beispielsweise des menschlichen Körpers gerichtet werden kann. Obwohl das von dem Halbleiter-Lasergenerator abgegebene Laserlicht auf den zu behandelnden Bereich beispielsweise über einen Spiegel übermittelt wird, ist es für den Zahnarzt schwierig, auf den zu behandelnden Bereich zu blicken, weil die Laserlicht-Bestrahlungsöffnung sehr nahe an dem Griff liegt. Wenn ferner Laserlicht auf eine innenliegende Stelle, beispielsweise einen Backenzahn im Mund des Patienten, gerichtet werden soll, kommt der Griff in störenden Eingriff mit den Wänden der Mundhöhle. Eine wirkungsvolle Behandlung ist ausgeschlossen.

Der Erfindung liegt die Aufgabe zugrunde, ein medizinisches Laserhandstück zu schaffen, das die vorstehend geschilderten Mängel ausräumt und insbesondere eine genau gezielte Behandlung auch schwierig zu erreichender Stellen zuläßt.

Diese Aufgabe wird mit den Maßnahmen des Patentanspruchs 1 gelöst. Dabei ist mit dem einen Halbleiter-Lasergenerator aufnehmenden Kopf des Laserhandstücks eine Bestrahlungsdüse lösbar verbunden. Über den endseitigen Durchlaß der Bestrahlungsdüse kann

das Laserlicht auf die zu behandelnde Stelle, beispielsweise im Rahmen einer Dentalbehandlung, mit hoher Genauigkeit aufgebracht werden.

Bevorzugte weitere Ausgestaltungen der Erfindung ergeben sich aus den Unteransprüchen.

Bevorzugte Ausführungsbeispiele der Erfindung sind nachstehend anhand der Zeichnungen näher erläutert. Es zeigen:

Fig. 1 einen Längsschnitt durch ein medizinisches Laserhandstück gemäß einer ersten Ausführungsform der Erfindung,

Fig. 2 eine Draufsicht auf die Rückseite des Laserhandstücks nach **Fig. 1**,

Fig. 3 einen Längsschnitt durch die wesentlichen Teile eines Laserhandstücks gemäß einer abgewandelten Ausführungsform der Erfindung,

Fig. 4 einen Längsschnitt durch ein Laserhandstück gemäß einer weiteren Ausführungsform der Erfindung,

Fig. 5 eine Draufsicht auf die Rückseite des Laserhandstücks gemäß **Fig. 4**,

Fig. 6 einen Teilschnitt eines Laserhandstücks gemäß einer weiter abgewandelten Ausführungsform der Erfindung,

Fig. 7 in größerem Maßstab einen Schnitt durch den Bereich *A* der **Fig. 6**,

Fig. 8 eine auseinandergezogene perspektivische Darstellung des Bereichs *B* der **Fig. 7**,

Fig. 9 in größerem Maßstab einen Schnitt des Bereichs *C* der **Fig. 6**,

Fig. 10 eine auseinandergezogene perspektivische Darstellung des Bereichs *D* der **Fig. 7**, und

Fig. 11 einen Schnitt entlang der Linie *E-E* der **Fig. 9**.

Das in den **Fig. 1** und **2** veranschaulichte Handstück *H* weist einen Griffkörper 1, einen Griffkopf 2 und eine Bestrahlungsdüse 5 auf. Die Bestrahlungsdüse 5 ist mit dem Griffkopf 2 des langgestreckten Griffkörpers 1 lösbar verbunden. Die Bestrahlungsdüse ist gegenüber dem Griffkörper abgewinkelt, und ihre Achse schneidet die Achse des Griffkörpers 1. Am Ende der Bestrahlungsdüse 5 befindet sich ein nach innen gebogener Abschnitt 19, der es erlaubt, das Ende der Bestrahlungsdüse 5 auf eine zu behandelnde Stelle zu richten, die schwierig zu erreichen ist (beispielsweise die Innenseite eines Zahns). Wenn die betreffende Stelle unter Verwendung einer geraden Bestrahlungsdüse behandelt werden kann, ist es nicht notwendig, den gebogenen Abschnitt 19 vorzusehen.

Auf das untere Ende des Griffkörpers 1 ist eine Verbindungshülse *S* 1 aufgeschraubt. Die Verbindungshülse *S* 1 trägt einen von außen betätigbaren Stromversorgungsschalter *S*, und sie weist an ihrem unteren Ende ein Steckverbindungsmitglied 11 auf. Mit dem unteren Ende der Verbindungshülse *S* 1 ist ein Anschlußschlauch 3 verbunden, der ein (nicht dargestelltes) Steckverbindungsmitglied aufweist, das zu dem Steckverbindungsmitglied 11 paßt. Der Stromversorgungsschalter *S* ist mit einer Steuereinrichtung 21, zu der eine gedruckte Steuerleiterplatte 20 gehört, über eine Anschlußleitung 75 verbunden. Die Steuereinrichtung 21, ein Bestrahlungsschalter 10 und ein Halbleiter-Lasergenerator 7 stehen über Verbindungsleitungen 76 untereinander in Verbindung. Die Steuereinrichtung 21 ist der Einfachheit halber nur schematisch dargestellt. Wenn der Stromversorgungsschalter *S* eingeschaltet wird, wird dem Bestrahlungsschalter 10 über die Steuereinrichtung 21 elektrische Energie von einer nicht dargestellten Stromquelle aus zugeführt. Beim Einschalten des Bestrahlungsschalters 10 geht elektrische Energie an den Halbleiter-La-

sergenerator 7.

Der Bestrahlungsschalter 10 wird eingeschaltet, wenn der Kopf des Schalters durch Verschwenken eines Steuerhebels 9 niedergedrückt wird, der mit dem Griffkörper 1 über einen Zapfen 101 gelenkig verbunden ist und der über dem Schalter 10 sitzt. Der Bestrahlungsschalter 10 schaltet ab, wenn er durch Rückstellen des Steuerhebels 9 freigegeben wird.

Der Steuerhebel 9 ist beispielsweise mit einem Handstückgehäuse 4 gelenkig verbunden und ragt über dieses Gehäuse in der in **Fig. 1** veranschaulichten Weise vor. Dem Steuerhebel 9 kann eine Verriegelungseinrichtung 28 (vergleiche auch **Fig. 5**) zugeordnet sein, um den Steuerhebel, falls erforderlich, verriegeln zu können.

Zu der Verriegelungseinrichtung 28 gehört ein unter dem Steuerhebel 9 sitzender Schieber 103 mit einem Schlitz 102, durch den der Zapfen 101 hindurchgreift. Wenn sich der Zapfen 101 entsprechend **Fig. 1** am unteren Ende des Schlitzes 102 befindet, ist eine am Schieber 103 vorgesehene Verriegelungsklaue 104 außer Eingriff mit dem Steuerhebel 9. Wird der Schieber 103 durch Handbetätigung nach unten bewegt, bis der Zapfen 101 das obere Ende des Schlitzes 102 erreicht, kommt die Verriegelungsklaue 104 mit dem unteren Innenabschnitt des Steuerhebels 9 in Kontakt, wodurch ein Verschwenken des Steuerhebels 9 verhindert wird. Auf diese Weise verhindert die Verriegelungseinrichtung 28 ein Einschalten des Bestrahlungsschalters 10.

Der in dem Griffkopf 2 untergebrachte Halbleiter-Lasergenerator 7, bei dem es sich beispielsweise um einen AlGaInP- oder AlGaAs-Laser handeln kann, ist mit der Achse der mit dem Griffkopf 2 verbundenen Bestrahlungsdüse 5 in der in **Fig. 1** veranschaulichten Weise ausgerichtet. In der Bestrahlungsdüse 5 sitzt eine Laserlicht-Übertragungseinrichtung 8, beispielsweise in Form einer optischen Faser, um von dem Halbleiter-Lasergenerator 7 erzeugtes Laserlicht zu übermitteln. Innerhalb eines endseitigen Durchlasses 18 der Düse ist eine Kondensorlinse 6a angeordnet, um das übermittelte Laserlicht zu kondensieren. Innerhalb des Griffkopfes 2 steht eine Sammellinse 6, welche das von dem Halbleiter-Lasergenerator 7 erzeugte Laserlicht zu einem Laserstrahl bündelt, mit dem Basisende der in der Bestrahlungsdüse 5 untergebrachten Laserlicht-Übertragungseinrichtung 8 in Kontakt. Ein Zwischenraum ist zwischen der Sammellinse 6 und dem Halbleiter-Lasergenerator 7 vorgesehen.

Eine Anzeigeeinrichtung 12 befindet sich an der Oberseite des Griffkopfes 2 (**Fig. 1** und **2**). Wenn der Stromversorgungsschalter *S* eingeschaltet wird, leuchtet eine Lampe *R* in einem Abschnitt 12a der Anzeigeeinrichtung 12 auf, um erkennen zu lassen, daß das Handstück für eine Bestrahlung bereit ist. Wird der Bestrahlungsschalter 10 eingeschaltet, leuchtet eine Lampe *EXP* im Abschnitt 12a auf; sie zeigt an, daß eine Bestrahlung durchgeführt wird. Die Lampen eines Abschnitts 12b der Anzeigeeinrichtung 12 werden benutzt, um die Bestrahlungsdauer anzuzeigen. Die Lampe, welche der mittels eines Steuerart-Wählschalters 13 ausgewählten Bestrahlungsdauer entspricht, leuchtet zur Anzeige der gewählten Betriebsart auf. Wenn die in einem Zeitgeber voreingestellte Bestrahlungsdauer abgelaufen ist, wird die betreffende Lampe selbsttätig abgeschaltet. An einer geeigneten Stelle der Laserlichtstrecke im Griffkopf 2 sitzt ein Laserdetektor 7a. Das von diesem Detektor 7a erfaßte Laserbestrahlungssignal wird einem Summer 67 zugeführt. Der Summer 67 er-

zeugt ein beispielsweise aus zwei Tönen bestehendes Schallsignal, das den Bedienenden erkennen läßt, daß eine Laserbestrahlung erfolgt. Wenn der Stromversorgungsschalter *S* eingeschaltet wird, erzeugt der Summer 67 ein anderes Schallsignal, das dem Bedienenden den Bereitschaftszustand anzeigt.

Während bei der Ausführungsform nach den Fig. 1 und 2 der in dem Griffkopf 2 untergebrachte Halbleiter-Lasergenerator 7 auf der Achse der mit dem Griffkopf 2 verbundenen Bestrahlungsdüse 5 liegt, läßt sich der Halbleiter-Lasergenerator 7 auch in der in Fig. 3 veranschaulichten Weise anordnen. Bei dieser Ausführungsform sitzt der Halbleiter-Lasergenerator 7 in dem Griffkörper 1 an einer Stelle, wo die Achse des Halbleiter-Lasergenerators 7 die Achse der Bestrahlungsdüse 5 schneidet. Es ist ferner ein Licht reflektierendes optisches Bauteil 6*b* vorgesehen, das dem Halbleiter-Lasergenerator 7 derart zugewendet ist, daß es das von dem Lasergenerator 7 erzeugte Laserlicht auf das Eintritts-ende der Laserlicht-Übertragungseinrichtung 8 umlenkt. Das Laserlicht durchläuft die Übertragungseinrichtung 8 in der Bestrahlungsdüse 5 und tritt über die Kondensorlinse 6*a* aus, die innerhalb des endseitigen Durchlasses 18 der Düse sitzt.

Eine Beleuchtungsvorrichtung 23, beispielsweise in Form einer Halogenlampe oder einer Glühlampe, sitzt auf der Achse der mit dem Griffkopf 2 verbundenen Bestrahlungsdüse 5. Das optische Bauteil 6*b* läßt das von der Beleuchtungsvorrichtung 23 abgegebene Licht durch, so daß es zu der Übertragungseinrichtung 8 in der Bestrahlungsdüse 5 gelangt. Mit Hilfe des von der Beleuchtungsvorrichtung 23 abgegebenen Lichts kann daher die Behandlungsstelle genau angepeilt und während der Behandlung beobachtet werden.

Es versteht sich, daß die Beleuchtungsvorrichtung 23 auch unabhängig von der Laserbestrahlung benutzt werden kann. Statt, wie dargestellt, die Beleuchtungsvorrichtung 23 mit der Achse der Düse 5 auszurichten, lassen sich der Halbleiter-Lasergenerator 7 und die Beleuchtungsvorrichtung 23 auch an anderer Stelle innerhalb des Griffkörpers 1 unterbringen, wobei zur Lichtübermittlung in bekannter Weise Prismen oder andere optische Bauteile herangezogen werden können.

Die Bestrahlungsdüse 5 ist mit dem Griffkopf 2 über O-Ringe 16 drehbar und lösbar verbunden. Dabei ist eine Überwurfmutter 15, in welche die Bestrahlungsdüse 5 eingesetzt ist, auf einen Verbindungsabschnitt 14 des Griffkopfes 2 aufgeschraubt. Die Bestrahlungsdüse 5 wird von den O-Ringen 16 drehbar und lösbar abgestützt. Außerdem verhindern die O-Ringe 16 einen unbeabsichtigten Austritt von Laserlicht. Die Bestrahlungsdüse 5 kann abgenommen und gegen eine andere Düse ausgetauscht werden, um beispielsweise die Düse an die zu behandelnde Stelle und/oder deren Lage am Patienten anzupassen. Außerdem kann der endseitige Durchlaß 18 der Düse 5 durch Drehen der Düse 5 auf die Behandlungsstelle ausgerichtet werden, während die Düse 5 mit dem Griffkopf 2 verbunden bleibt.

Die Fig. 4 und 5 zeigen eine abgewandelte Ausführungsform des Handstücks *H*, bei welcher Zuleitungen 24, 25, 26 und 27 im Griffkörper 1 und in der Bestrahlungsdüse 5 vorgesehen sind, um Arbeitsmedien, beispielsweise Luft *A* und Wasser *W*, von dem am unteren Ende der Verbindungshülse *S* 1 angeschlossenen Schlauch 3 aus dem Griffkörper 1 und der Bestrahlungsdüse 5 zuzuführen. In diesem Fall kann der mit dem Griffkörper 1 gelenkig verbundene Hebel 9 in einen Luftsteuerhebel 9*a* und einen Wassersteuerhebel 9*b*

aufgeteilt werden, wie dies in Fig. 5 dargestellt ist. Auf diese Weise können die Arbeitsmedien *A* und *W* von dem mit dem unteren Ende des Griffkörpers verbundenen Schlauch 3 aus dem endseitigen Düsendurchlaß 18 zugeführt oder abgesperrt werden, indem der entsprechende Steuerhebel verschwenkt wird.

Wie aus Fig. 4 hervorgeht, sind in die Verbindungshülse *S* 1 eine Luftzuleitung 240 und eine hinter dieser Luftzuleitung liegende Wasserzuleitung (nicht veranschaulicht) eingesetzt. Diese Zuleitungen sind an die Luftzuleitung 24 und die Wasserzuleitung 25 in dem Griffkörper 1 über eine Ventilanordnung 68 angeschlossen, die über den Steuerhebel 9 betätigbar ist. Die Zuleitungen der Verbindungshülse stehen ferner mit der Luftzuleitung 26 und der Wasserzuleitung 27 in Verbindung, die beide in die Bestrahlungsdüse 5 eingebaut sind und die sich in Richtung der Achse dieser Düse erstrecken. Die Wasserzuleitung 27 reicht als einzelnes Rohr in der Bestrahlungsdüse 5 bis zu dem endseitigen Durchlaß 18. Die Luftzuleitung 26 erstreckt sich über fast die gesamte Länge der Düse 5, und sie verzweigt sich konzentrisch zu der Außenwand der Zuleitung 27. Luft und Wasser, die über die Zuleitungen 26 und 27 zugeführt werden, werden am Ende der Düse 5 unter Bildung eines Nebels teilweise miteinander kombiniert. Der so erzeugte Nebel kann von dem endseitigen Durchlaß 18 abgegeben werden. Die im einzelnen vorgesehenen Mittel zur Erzeugung des Nebels sind nicht dargestellt.

Die Ventilanordnung 68 wird durch Verschwenken des Steuerhebels 9 geöffnet oder geschlossen. Eine Ventilkugel 682 verschließt unter dem Einfluß einer Feder 681 eine Ventilöffnung 683. Wenn der Steuerhebel 9 niedergedrückt wird, wird eine Ventilstange 684 nach unten geschoben. Dadurch wird die Ventilkugel 682 abgesenkt; das Ventil öffnet. Infolgedessen gelangt Luft über einen Verbindungskanal 243 zu der Luftzuleitung 24. Die Wasserzufuhr kann in gleicher Weise wie die Luftzufuhr gesteuert werden. Abweichend von den Ausführungsformen nach den Fig. 1, 2 und 3 sitzt in diesem Fall der Bestrahlungsschalter 10 an der Innenseite des Griffkörpers 1. Der Laserdetektor 7*a* befindet sich an einer von den Zuleitungen 24 und 25 abliegenden Stelle. Wenn dem Griffkörper 1 durch Betätigen des Stromversorgungsschalters *S* des Laserhandstücks *H* elektrische Energie von einer (nicht dargestellten) Stromquelle aus zugeführt wird, gibt der Summer ein Schallsignal ab, das anzeigt, daß sich das Handstück *H* im Bereitschaftszustand befindet. Wenn der im Griffkopf 2 angeordnete Zeitgeber-Wählschalter 13 auf eine dem jeweiligen Behandlungszweck entsprechende Betriebsart eingestellt wird, wird die gewählte Betriebsart durch die Zeitgeber-Anzeigelampe 12*b* der Anzeigeeinrichtung 12 angezeigt.

Die Bestrahlungsdüse 5 kann gegenüber dem Griffkopf 2 gedreht werden, um den endseitigen Durchlaß 18 mit der zu behandelnden Stelle auszurichten. Wird dann der Bestrahlungsschalter 10 gedrückt, ändert sich das vom Summer abgegebene Schallsignal, beispielsweise von einem Dauerton zu einem intermittierenden Ton, um anzuzeigen, daß der Halbleiter-Lasergenerator 7 aktiviert ist und Laserlicht abgegeben wird. Die Pilotlampe 12*a* der Anzeigeeinrichtung 12 wird durch ein Erfassungssignal des Laserdetektors 7*a* eingeschaltet.

Das von dem Halbleiter-Lasergenerator 7 erzeugte Laserlicht wird mittels der Sammellinse 6 zu einem engen Strahl gebündelt. Dieser Strahl wird über die in der Bestrahlungsdüse 5 untergebrachte Laserlicht-Übertragungseinrichtung 8 weitergeleitet und dann mittels der

Kondensorlinse 6a weiter gebündelt, bevor er auf die zu behandelnde Stelle trifft. Weil das von dem Lasergenerator 7 kommende Laserlicht über die mit dem Griffkopf 2 verbundene und gegenüber dem Griffkörper abgewinkelte Bestrahlungsdüse 5 abgestrahlt wird, kann der Bedienende das Laserlicht auf die jeweils zu behandelnde Stelle richten, während er den Griffkörper 1 faßt, ohne dabei eine unnatürliche Haltung einnehmen zu müssen. Die Behandlung läßt sich infolgedessen auch über längere Zeiträume hinweg ermüdungsfrei durchführen. Das Feld 12c der Anzeigeeinrichtung 12 ist dem Bedienenden während der Behandlung zugewendet, so daß der Bedienende die Anzeige überprüfen kann, ohne die Haltung für die Behandlung ändern zu müssen. Außerdem wird der Bedienende durch den von dem Laserdetektor 7a gesteuerten Summer davon unterrichtet, ob Laserlicht abgegeben wird oder nicht. Die Behandlung kann daher aufgrund der optischen und akustischen Anzeigesignale besonders sicher durchgeführt werden.

Bei der Ausführungsform nach den Fig. 1, 2 und 3 wird der Bestrahlungsschalter 10 automatisch ausgeschaltet, wenn der Bedienende den Steuerhebel 9 freigibt. Im Falle der Ausführungsform nach den Fig. 4 und 5 wird die Laserbestrahlung unterbrochen und das Handstück in den Bereitschaftsbetrieb zurückgebracht, indem der Bestrahlungsschalter 10 ausgeschaltet wird.

Wenn entsprechend Fig. 3 eine Laserlicht-Übertragungseinrichtung, beispielsweise in Form einer optischen Faser, im Griffkopf 2 entlang der Achse der Bestrahlungsdüse 5 vorgesehen wird und wenn die Beleuchtungsvorrichtung, beispielsweise in Form einer Halogenlampe, an einer Stelle sitzt, an der sich bei der Ausführungsform nach Fig. 1 der Lasergenerator 7 befindet, kann das unsichtbare Laserlicht mit Hilfe des von der Beleuchtungsvorrichtung 23 abgegebenen Lichts verfolgt und gerichtet werden, was die Behandlung besonders sicher und wirkungsvoll macht. Wird die Beleuchtungsvorrichtung 23 unabhängig von einer Laserbestrahlung genutzt, kann beispielsweise der für eine Behandlung vorgesehene Bereich optisch überprüft werden.

Das in den Fig. 4 und 5 veranschaulichte Handstück H, bei dem die Zuleitungen 24, 25, 26 und 27 für Luft A und Wasser W in dem Griffkörper 1 und der Bestrahlungsdüse 5 untergebracht sind, ermöglicht vier unterschiedliche Betriebsarten, nämlich: Laserbestrahlung mittels des Halbleiter-Lasergenerators 7, Aufblasen eines Luftstrahls A, Aufblasen eines Wasserstrahls W und Erzeugung eines Sprühnebels durch Mischen von Luft und Wasser. Das Handstück H ist infolgedessen als medizinisches Handstück vielseitig einsetzbar.

Bei den vorstehend erläuterten medizinischen Laserhandstücken sitzt der Halbleiter-Lasergenerator im Griffkopf, und die Bestrahlungsdüse ist mit dem Griffkopf unter Bildung eines Winkels verbunden, so daß Laserlicht über die Übertragungseinrichtung innerhalb der Bestrahlungsdüse zu jeder beliebigen Behandlungsstelle geleitet werden kann. Der Bedienende kann daher die Behandlungsstelle beobachten und Laserlicht genau auf die Behandlungsstelle auftreffen lassen, ohne daß während der Behandlung der Griffkörper zu nahe an die Behandlungsstelle, beispielsweise innerhalb des Mundes des Patienten, herangebracht zu werden braucht. Die Ausrichtung und Lage des endseitigen Durchlasses der Bestrahlungsdüse läßt sich bedarfsweise ändern, indem die Bestrahlungsdüse gegenüber dem Griffkopf gedreht wird. Die Bestrahlungsdüse kann ferner abgenommen und leicht gegen eine neue und/oder andere Bestrah-

lungsdüse ausgetauscht werden. Wenn die Bestrahlungsdüse nicht verwendet wird, kann man Laserlicht auf einen relativ weiten Bereich, beispielsweise die Frontzähne, auffallen lassen. Das Handstück erlaubt es, Zahn- und Periodontalbereiche im Mund, die normalerweise schwierig zu behandeln sind, weil sie relativ klein sind und komplizierte Form haben, problemlos und einfach zu behandeln. Der Wirkungsgrad der Behandlung läßt sich dadurch erheblich steigern.

Bei dem Handstück H gemäß Fig. 6 werden dem Griffkörper 1 und der Bestrahlungsdüse 5 Luft und Wasser in ähnlicher Weise zugeführt wie bei dem Ausführungsbeispiel nach den Fig. 4 und 5. Auch in diesem Fall kann ein Nebel aus Luft und Wasser gebildet werden. Der Bestrahlungsschalter 10, der Steuerhebel 9 und weitere im Griffkörper 1 untergebrachte Bauteile sind in Fig. 6 der Einfachheit halber nicht dargestellt. Abweichungen von den zuvor erläuterten Ausführungsformen sind nachstehend unter Bezugnahme auf die Fig. 6 bis 11 beschrieben.

Die Luftzuleitung 24 und die Wasserzuleitung 25 reichen vom Griffkopf 2 zu der Bestrahlungsdüse 5. In dem Griffkörper 1 ist die gedruckte Steuerleiterplatte 20 zum Steuern des Halbleiter-Lasergenerators 7 untergebracht. Die Steuerleiterplatte 20 steht mit dem im Griffkopf 2 sitzenden Lasergenerator 7 in elektrischer Verbindung.

Ein rohrförmiges Innengehäuse 31 ist im Verbindungsbereich von Griffkopf 2 und Bestrahlungsdüse 5 koaxial zu letzterer eingebaut (Fig. 7). An dem Basisendabschnitt des Innengehäuses 31 ist der Halbleiter-Lasergenerator 7 über eine Scheibe 32 mittels eines mit Gewinde versehenen Halteteils 29 befestigt, das an seiner hinteren Endfläche Kühlrippen 30 trägt. In das Innengehäuse 31 sind die Luftzuleitung 24 und die Wasserzuleitung 25 eingebaut. Außerdem zweigt von der Luftzuleitung 24 ein Luftdurchlaß 34 kleinen Durchmessers ab, um den Halbleiter-Lasergenerator 7 zu kühlen. Im Lasergenerator-Einbaubereich des Innengehäuses 31 sind Schlitze 35 ausgebildet (Fig. 8). Mehrere Nuten 36 befinden sich in der mit dem Halbleiter-Lasergenerator 7 in Kontakt stehenden Seite der Scheibe 32. Die über den Luftdurchlaß 34 zugeführte Kühlluft tritt durch die Schlitze 35 und die Nuten 36 um die Rückseite des Lasergenerators 7 herum ein. Ein dem Kühlen des Lasergenerators 7 dienender Zwischenraum 37 wird von den Schlitzen 35 und den Nuten 36 gebildet. Die Scheibe 32 weist kleine durchgehende Öffnungen 38 zur Aufnahme der elektrischen Zuleitungen des Halbleiter-Lasergenerators 7 auf.

Am Basisendbereich des Innengehäuses 31 befindet sich ein Laserlicht-Aufnahmeabschnitt 39, dessen Durchmesser nach vorne allmählich abnimmt. Die sich verjüngende Oberfläche dieses Abschnittes ist mit Gold beschichtet oder mit einem Spiegelfinish versehen, um für eine Totalreflektion zu sorgen, so daß das von dem Halbleiter-Lasergenerator 7 kommende Laserlicht verlustfrei übermittelt werden kann. Benachbart dem Laserlicht-Aufnahmeabschnitt 39 sind optische Linsen 40 eingebaut. Diese Linsen werden von Halteteilen 41 und 42 gehalten.

Die Bestrahlungsdüse 5 besteht im wesentlichen aus einem Außenmantel 43, der in den Außenmantel eingesetzt, sich entlang dessen Achse erstreckenden Laserlicht-Übertragungseinrichtung 8, beispielsweise in Form einer Glasfaser, und einem flexiblen Schutzrohr 81 zum Schutz der Übertragungseinrichtung 8 (Fig. 9). Ein rohrförmiges Basisteil 80 ist in das Innengehäuse 31 des

Kopfs 2 über eine Steuerhülse 44 lösbar eingesetzt. Wie aus den Fig. 7 und 10 hervorgeht, befinden sich am Basisteil 80 ein Kupplungsstück 45 für das Schutzrohr 81, Halteteile 65 und 66 zur Verbindung des Kupplungsstücks mit der Glasfaser 8 und ein Spannglied 46 zur Verbindung des Kupplungsstücks 45 mit dem Außenmantel 43. Das Halteteil 65 dient gleichzeitig dem Übermitteln von Laserlicht, das von den Linsen 40 kommt, zu der Glasfaser 8. Der Laserlicht aufnehmende Abschnitt des Halteteils 65 ist mit Gold beschichtet oder mit einem Spiegelfinish versehen, und er verjüngt sich nach vorne. Über die Wasserzuleitung 25 zugeführtes Wasser tritt durch eine Umfangsnut 47 im Verbindungsbereich zwischen dem Griffkopf 2 und der Bestrahlungsdüse 5 sowie durch Durchgangsöffnungen 49 und 50 in dem Spannglied 46 und dem Kupplungsstück 45 hindurch. Das Wasser gelangt in einen Zwischenraum 56 zwischen der Glasfaser 8 und dem Schutzrohr 81, und es verläßt strahlförmig das Ende der Düse, während es die Außenfläche der Glasfaser 8 kühlt. Die über die Luftzuleitung 24 zugeleitete Luft durchströmt eine Umfangsnut 53 und eine Durchgangsöffnung 54 in dem Spannglied 46. Die Luft gelangt in einen Zwischenraum 55 zwischen dem Schutzrohr 81 und dem Außenmantel 43, und sie verläßt in Form eines Strahls das Ende der Düse 5, während sie die Außenfläche des Schutzrohrs 81 kühlt.

Die Glasfaser 8 und das Schutzrohr 81 sind in dem Außenmantel 43 koaxial zueinander abgestützt, wobei die Zwischenräume 56 und 55 gebildet werden. Am Ende der Glasfaser 8 ist ein Kupplungsstück 58 mit mehreren nach innen ragenden Vorsprüngen 57 aufgesetzt und mit dem Schutzrohr 81 verbunden (Fig. 9 und 11). Die Vorsprünge 57 legen sich gegen die Umfangsfläche der Glasfaser 8 an, um den Zwischenraum 56 zwischen der Glasfaser 8 und dem Schutzrohr 81 auszubilden. Kühlwasser tritt über die Zwischenräume zwischen den Vorsprüngen 57 aus. Um das Kupplungsstück 58 herum ist eine Endkappe 60 auf den Außenmantel 53 unter Zwischenfügen eines O-Rings 59 aufgeschraubt. Zwischen der Endkappe 60 und dem Kupplungsstück 58 ist ein Abstandshalter 61 eingesetzt, der für einen Luftauslaßkanal sorgt. Vorsprünge 62 sind in geeigneten Abständen an der Innenumfangsfläche des Endabschnitts des Abstandshalters 61 vorgesehen. Diese Vorsprünge 62 legen sich gegen die Außenumfangsfläche des Endabschnitts des Kupplungsstücks 58 an. Ein gefräster Abschnitt 63 ist um die Außenumfangsfläche des Kupplungsstücks 58 herum ausgebildet, und zwischen dem Abschnitt 63 und den Vorsprüngen 62 des Abstandshalters 61 liegende Zwischenräume 621 führen zu dem Zwischenraum 55. Luft und Wasser, die über die Luftzuleitung 24 und die Wasserzuleitung 25 zugeführt werden, kühlen den Halbleiter-Lasergenerator 7, die Glasfaser 8 und das Schutzrohr 81; sie verlassen in Strahlform das Ende der Bestrahlungsdüse 5, um die zu behandelnde Stelle zu kühlen oder andere für die Behandlung notwendige Funktionen zu erfüllen. Steuerventile für die Arbeitsmedien (Luft und Wasser) befinden sich am Griffkörper 1 oder an anderer von der Düse 5 abliegender Stelle, so daß der Bedienende die Ventile steuern kann, um ähnlich wie bei einer konventionellen Dreiwege-Dentalspritze Luft, Wasser oder einen Nebel aus Luft und Wasser vom Ende der Düse 5 aus abzugeben. O-Ringe 64 sorgen für luft- und wasserdichte Verbindungen. Die Form des Zwischenraums 37 ist nicht auf die in den Figuren dargestellte Ausführungsform beschränkt. Die Erfindung ist auch bei Handstücken von anderer Form anwendbar.

Bei dem vorstehend erläuterten Laserhandstück wird die Umfangsfläche des Halbleiter-Lasergenerators luftgekühlt, und das Lasergenerator-Halteteil ist mit Kühlrippen versehen. Dadurch wird verhindert, daß sich der Halbleiter-Lasergenerator übermäßig aufheizt. Ausgangsverluste des Generators werden herabgesetzt, obwohl der Generator in einen so begrenzten Raum wie das Innere eines Dentalhandstücks eingebaut ist. Die Ausgangsleistung des Halbleiter-Lasergenerators kann entsprechend erhöht werden.

Die Behandlung mit Laserlicht kann, wie erläutert, durch die Zufuhr von Wasser, Luft oder einem Nebel aus Luft und Wasser ergänzt werden. Durch Aktivieren der Zufuhr von Arbeitsmedium während der Laserlichtbehandlung kann die behandelte Stelle gekühlt werden. Die Stimulationswirkung des Laserlichts läßt sich steigern. Die behandelte Stelle und der umliegende Bereich werden gereinigt. Wenn ein Sprühnebel abgegeben wird, umschließt der Luftstrahl den Wasserstrahl, um für eine gleichförmige Nebelbildung zu sorgen. Die Kühl- und Reinigungsfunktionen des Handstückes werden weiter gesteigert. Außerdem ist die Laser-Übertragungseinrichtung von Wasser umgeben, wodurch sie gekühlt wird und Energieverluste während der Laserlichtübermittlung verringert werden. Das Handstück kann daneben auch als konventionelle Dreiwegespritze eingesetzt werden, wenn nicht mit Laserlicht gearbeitet wird.

Wie gleichfalls oben erläutert ist, kann der lichtaufnehmende Teil zwischen dem Halbleiter-Lasergenerator und der Laserlicht-Übertragungseinrichtung mit einer totalreflektierenden verjüngten Oberfläche ausgestattet sein. Das von dem Halbleiter-Lasergenerator erzeugte Laserlicht wird dabei durch die sich verjüngende Oberfläche totalreflektiert und in Vorwärtsrichtung konvergiert. Nahezu 100% des Laserlichts können in die Laserstrahl-Übertragungseinrichtung eintreten. Dadurch werden Laserlichtverluste vermindert. Es wird für eine besonders wirkungsvolle Behandlung unter voller Ausnutzung der Ausgangsleistung des Lasergenerators gesorgt. Es kommt nicht zu Wärmeerzeugung durch ausleckendes Laserlicht. Um das Handstück herum verwendete Instrumente werden durch Wärmeerzeugung aufgrund von ausleckendem Laserlicht nicht nachteilig beeinflusst.

Vorzugsweise sind die oben erläuterten weiteren Ausgestaltungen in Kombination vorgesehen. Sie können aber auch unabhängig voneinander angewendet werden.

Offenlegungstag:

21. Januar 1988

FIG. 1

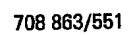
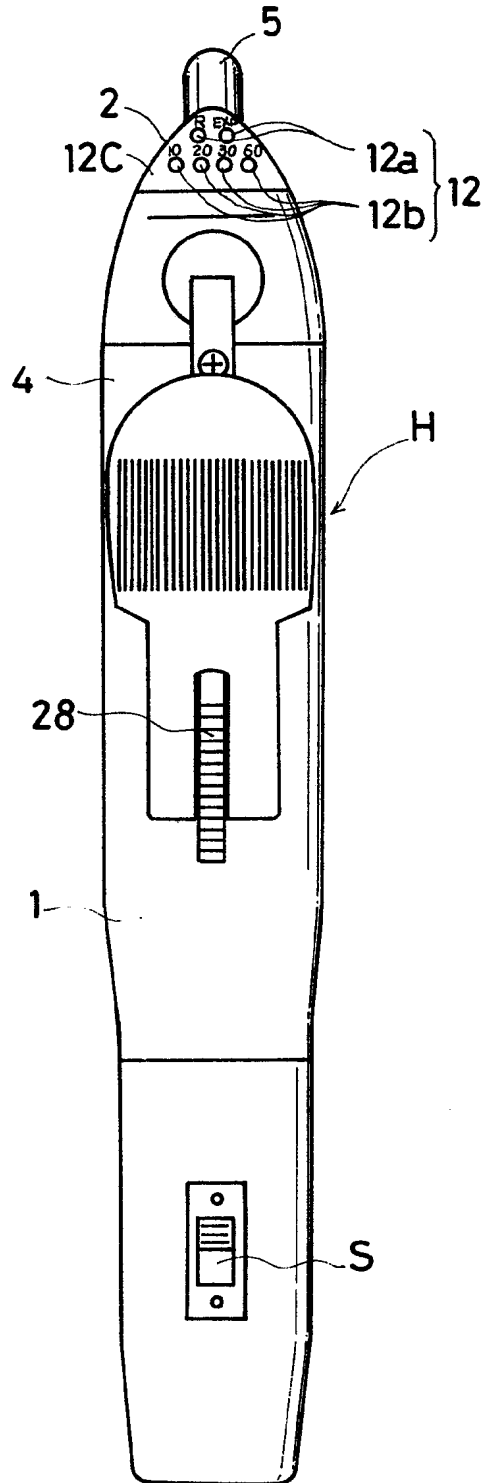


FIG. 2

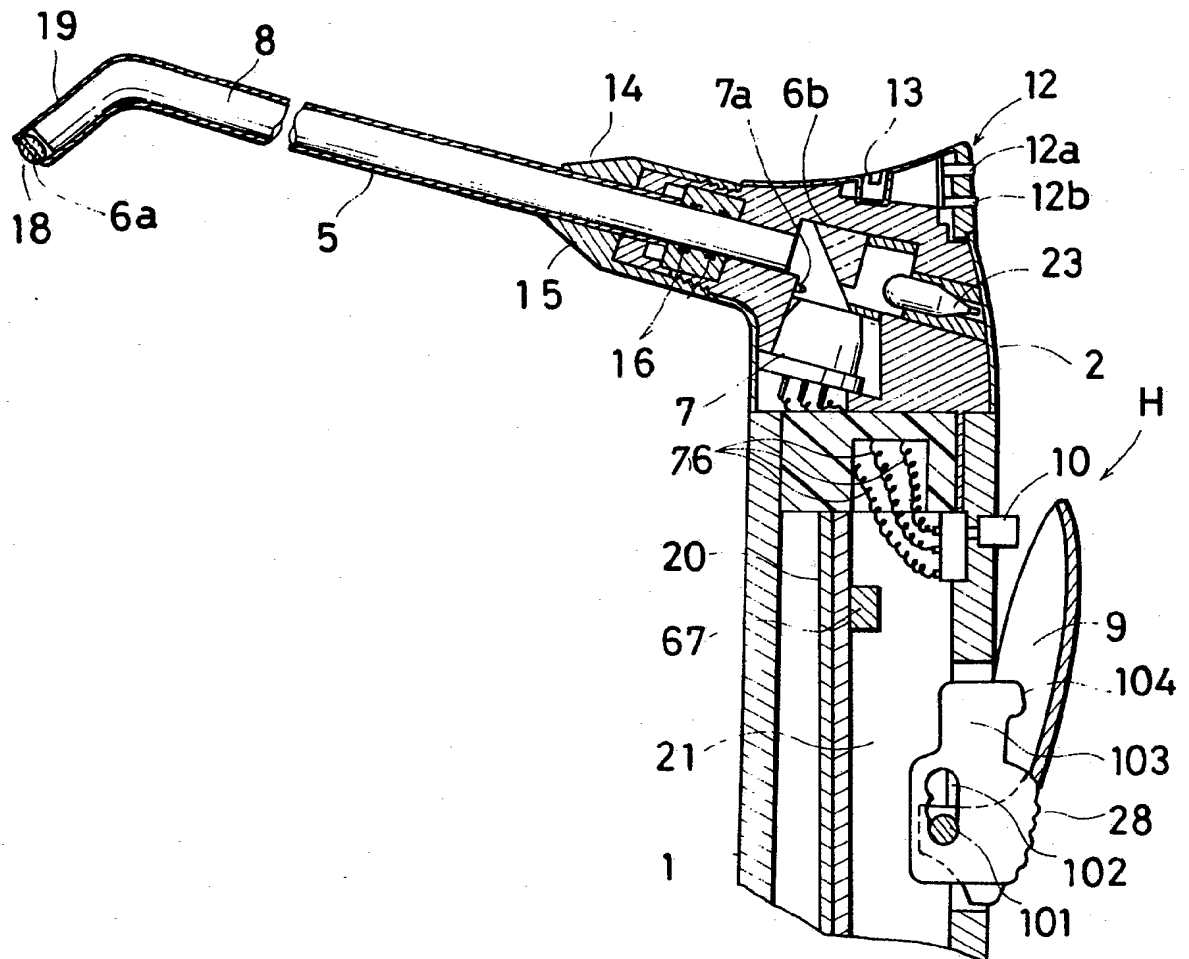
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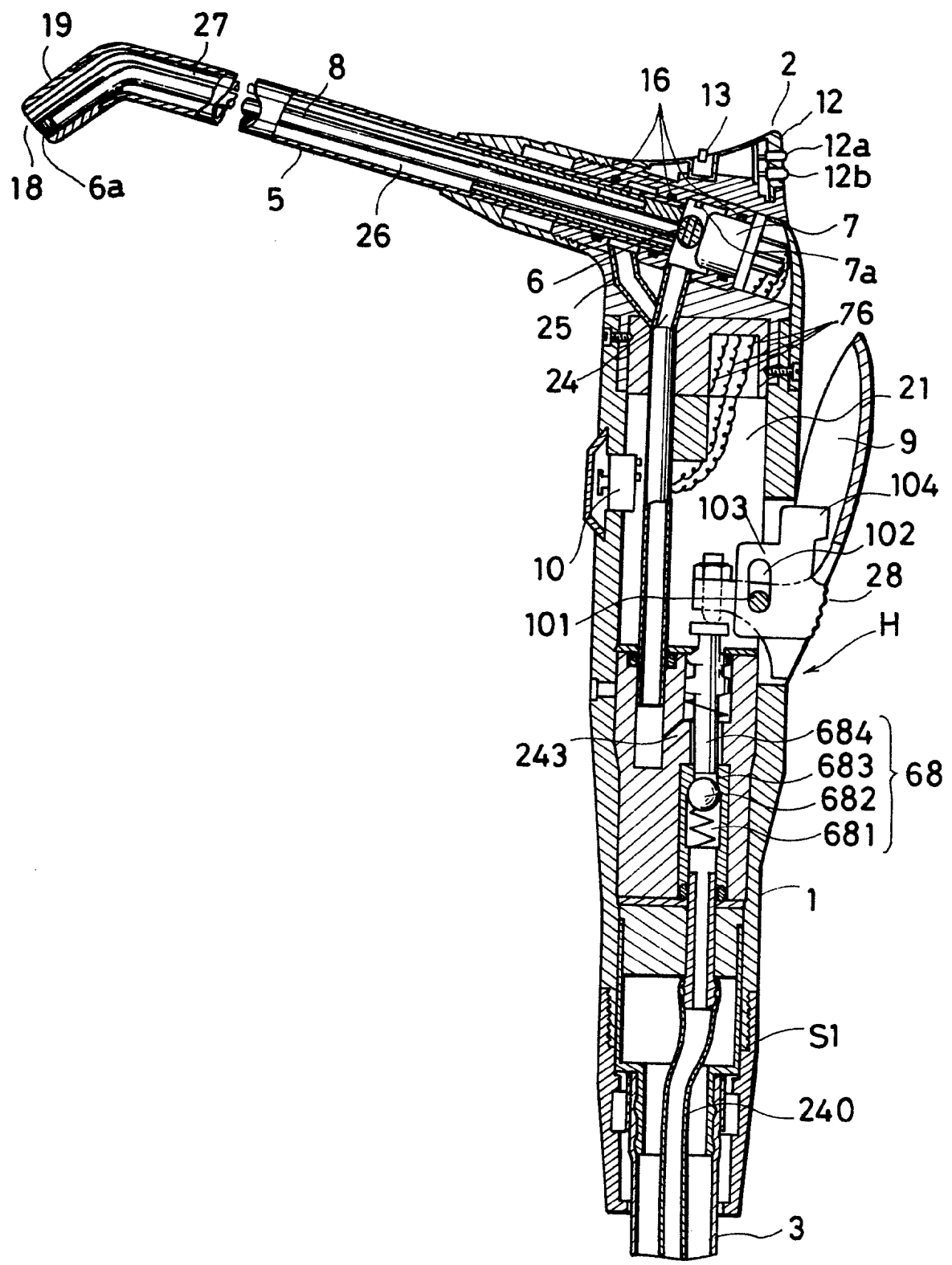
FIG. 3



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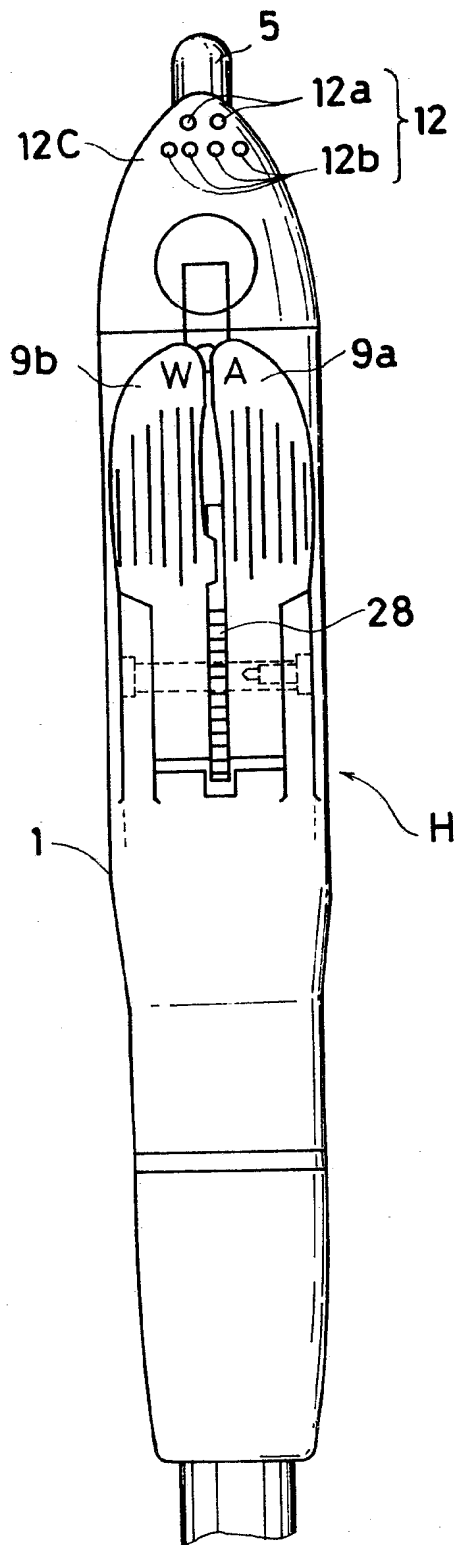
FIG. 4



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FIG. 5



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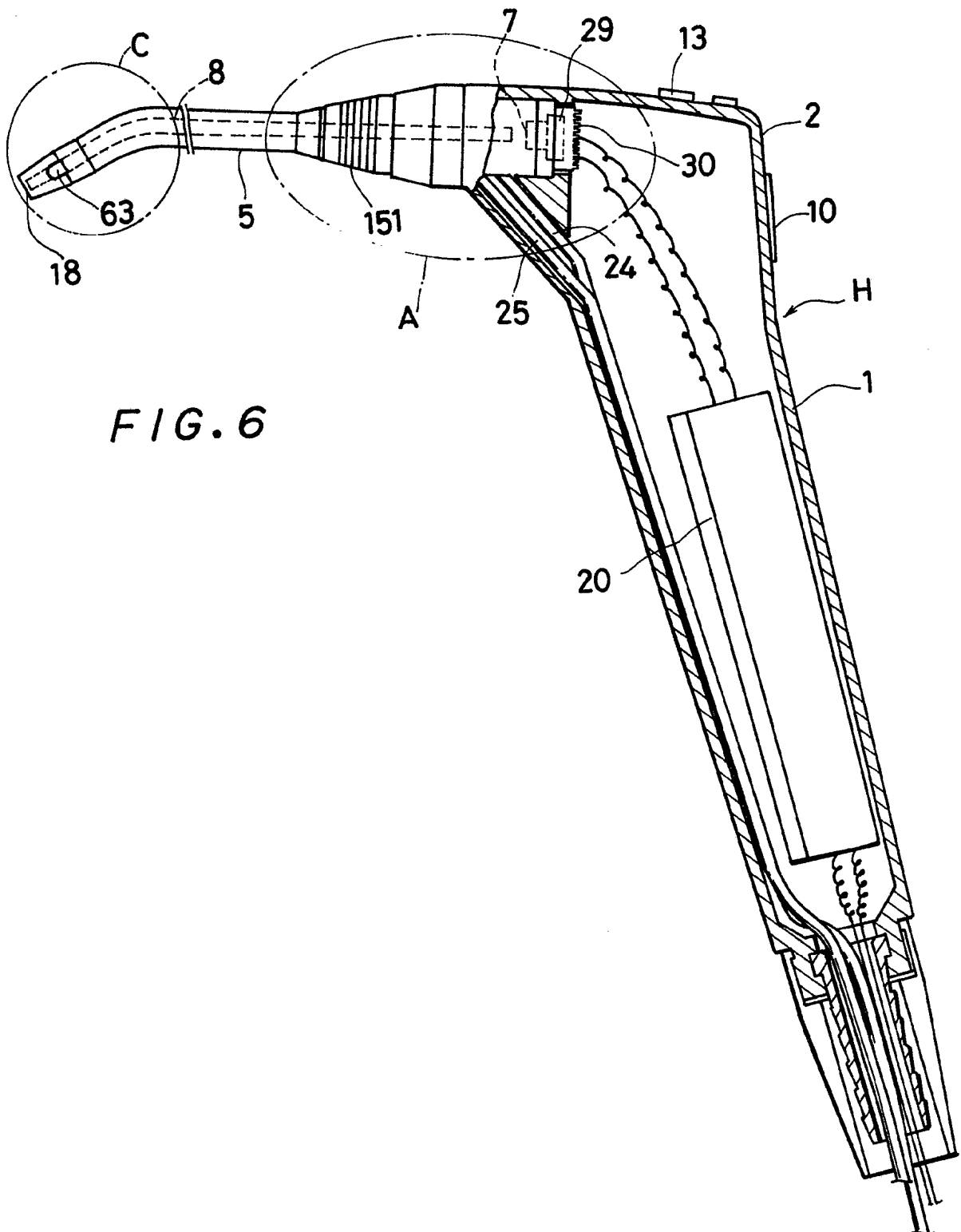
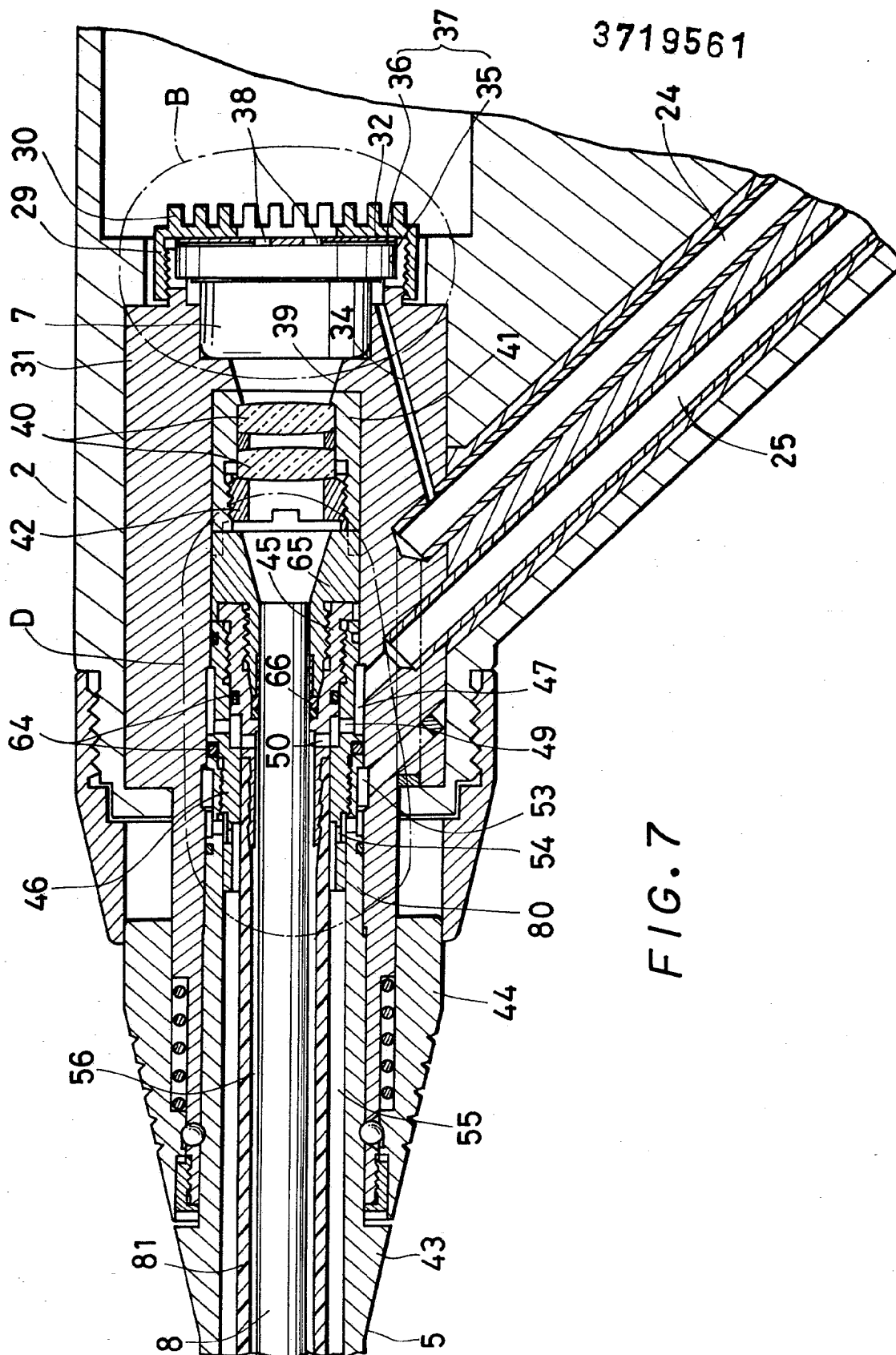
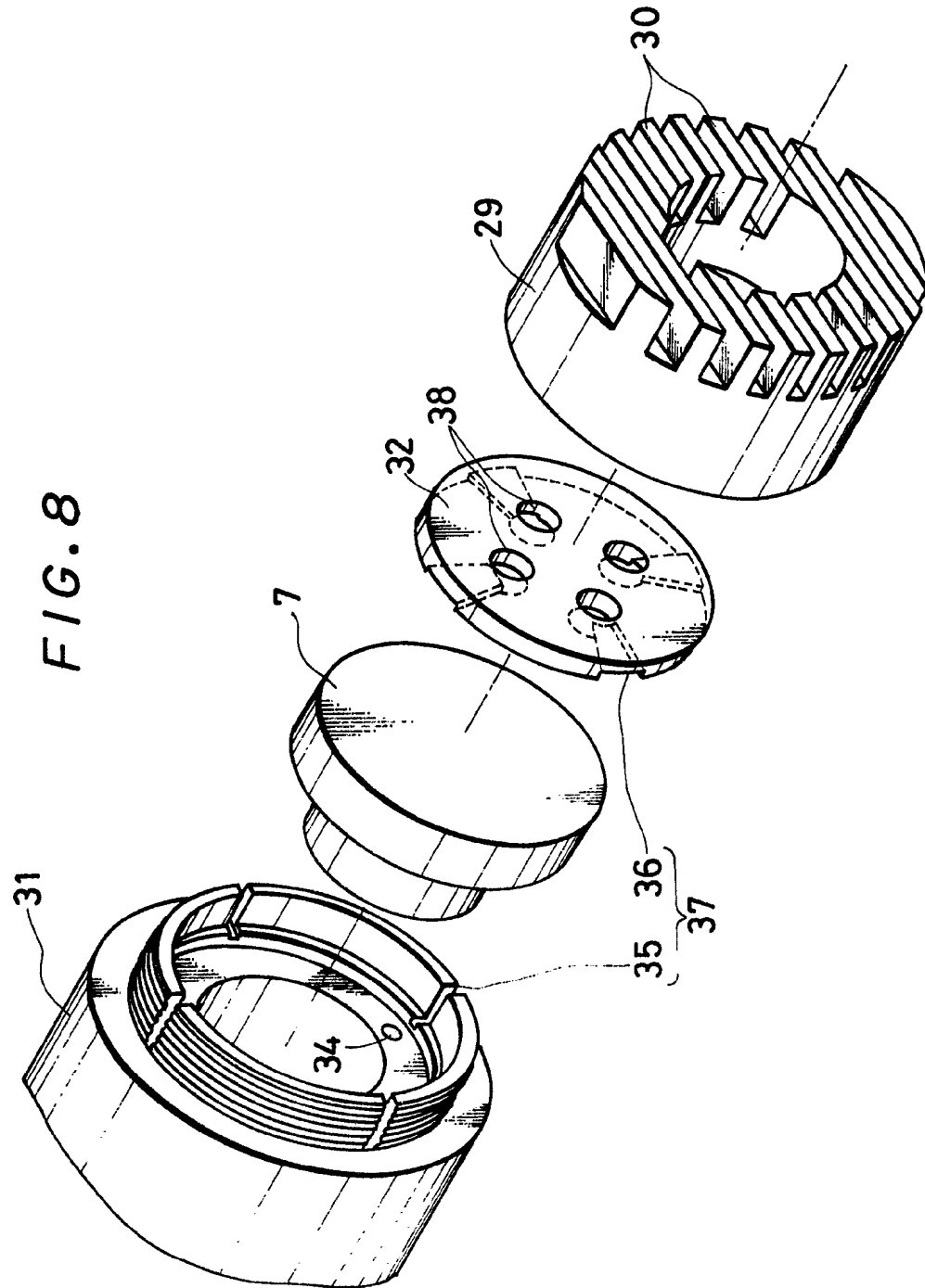


FIG. 6

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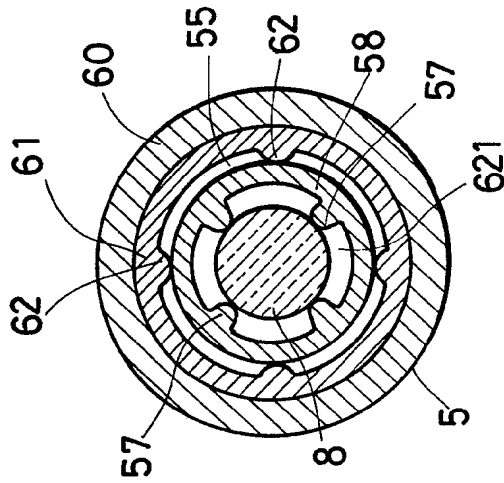
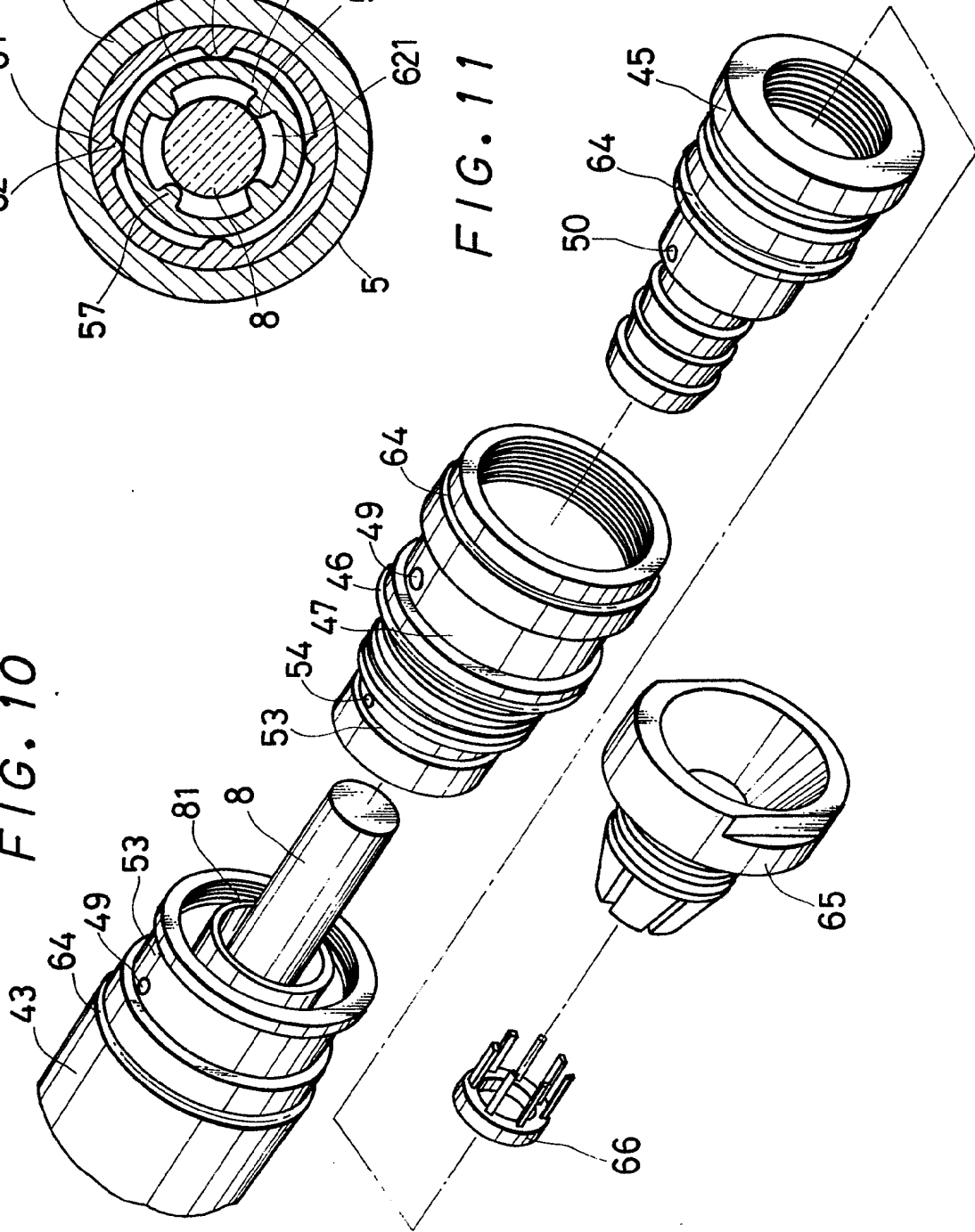


FIG. 11

FIG. 10



Device for treating skin lesions

Publication number: DE3837248 (A1)

Publication date: 1990-05-03

Inventor(s): TEICHMANN HEINRICH OTTO DR [DE]; TEICHMANN HARRO DR MED [DE]

Applicant(s): TEICHMANN HEINRICH OTTO DR PHY [DE]; TEICHMANN HARRO DR MED [DE]

Classification:

- **international:** **A61B18/20**; A61B17/00; **A61B18/20**; A61B17/00; (IPC1-7): A61N5/06; G06F15/64; H04N7/18

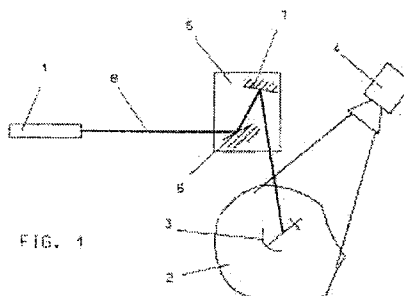
- **European:** A61B18/20H

Application number: DE19883837248 19881028

Priority number(s): DE19883837248 19881028

Abstract of DE 3837248 (A1)

There is specified a device for treating skin lesions on the human or animal body with the aid of a laser beam in which an image recording system (4) is provided for recording and storing the images of a skin area (2) to be recorded and space coordinates for the area to be treated are derived from the stored image by a discriminator. This can be used to derive a sequence of commands for control of the impact coordinates of the laser beam on the area to be treated. The invention can be used in all types of skin lesions, e.g. moles, tattoos, warts and the like.



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54 Einrichtung zur Behandlung von Hautveränderungen

Es wird eine Einrichtung zur Behandlung von Hautveränderungen des menschlichen oder tierischen Körpers mit Hilfe eines Laserstrahls angegeben, bei der ein Bilderfassungssystem (4) zur Aufnahme und Speicherung der Bilder eines zu erfassenden Hautgebietes (2) vorgesehen ist und durch einen Diskriminator aus dem gespeicherten Bild Raumkoordinaten für das zu behandelnde Gebiet abgeleitet werden. Daraus läßt sich eine Befehlsfolge zur Steuerung der Auftreffkoordinaten des Laserstrahls auf das zu behandelnde Gebiet ableiten. Die Erfindung ist für alle Arten von Hautveränderungen, wie Feuermale, Tätowierungen, Warzen usw. verwendbar.

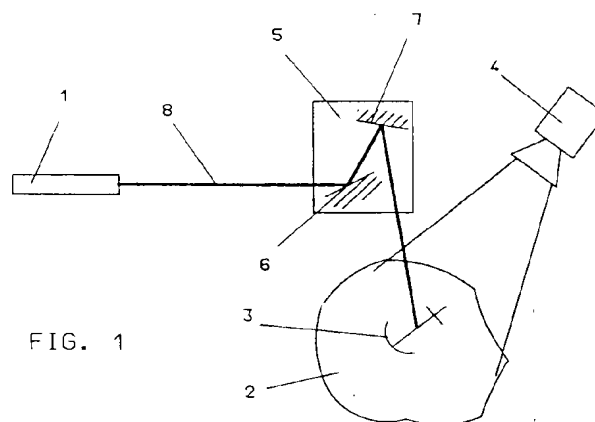


FIG. 1

Die Erfindung betrifft eine Einrichtung zur Behandlung von Hautveränderungen des menschlichen oder tierischen Körpers nach dem Oberbegriff des Anspruchs 1.

Seit einigen Jahren ist es bekannt, Hautveränderungen, wie z. B. Tätowierungen, Warzen, Feuermale oder ähnliches durch einen Laserstrahl zu behandeln. Der Laserstrahl erlaubt die schonende und blutarme Behandlung von Pigment- oder Strukturveränderungen der Haut. Dazu werden u.a. Helium-Neon, Argon- oder CO₂-Laser verwendet.

Üblicherweise wird der Laserstrahl von Hand über das zu behandelnde Hautgebiet geführt. Nicht nur bei flächenhafter Anwendung kommt es dabei unweigerlich zu erheblichen Schwankungen der Energieeinwirkung des Lasers. Dies führt teils zu einer vermehrten Narbenbildung, teils zu nicht ausreichend bestrahlten Gebieten. Eine exakte Dosierbarkeit der Energiemenge, die in ein Volumenelement der Haut eingebracht wird, ist daher auf diese Weise nicht möglich.

Es ist zwar auch möglich, einen Laserstrahl selbsttätig über ein Behandlungsgebiet, z. B. mit konstanter Geschwindigkeit, zu führen, wodurch eine verbesserte Dosierbarkeit erreichbar ist. Allerdings treten dabei erhebliche Probleme an Behandlungsgrenzen auf, denn der Laserstrahl soll nur das tatsächlich zu behandelnde Gebiet erfassen.

Der Erfindung liegt die Aufgabe zugrunde, eine Einrichtung zur Behandlung von Hautveränderungen des menschlichen oder tierischen Körpers mit Hilfe eines Laserstrahls anzugeben, die eine exakte Dosierbarkeit der Energieeinbringung in die Haut über das gesamte zu behandelnde Gebiet ermöglicht, ohne daß angrenzende gesunde Haut angegriffen wird.

Diese Aufgabe wird durch die in Anspruch 1 angegebene Erfindung gelöst. Vorteilhafte Weiterbildungen der Erfindung sind in Unteransprüchen angegeben.

Gemäß der Erfindung ist ein Bilderfassungssystem zur Aufnahme und Speicherung der Bilder eines zu behandelnden Hautgebietes vorgesehen. Aus einem oder mehreren gespeicherten Bildern werden über einen Diskriminator mit einstellbarer Entscheidungsschwelle Raumkoordinaten des zu behandelnden Gebietes abgeleitet. Aus den derart gewonnenen Raumkoordinaten wird eine Befehlsfolge abgeleitet, die zur Steuerung der Auftreffkoordinaten des Laserstrahls auf dem zu behandelnden Gebiet dient.

Die Bilderfassung des zu behandelnden Hautgebietes kann z. B. über eine TV-Kamera erfolgen, wobei vorzugsweise mehrere Bilder in verschiedenen Spektralbereichen aufgezeichnet werden. Durch einen Bildvergleich der gespeicherten Aufnahmen bei wählbarer Diskriminatorsschwelle lassen sich die Raumkoordinaten des zu behandelnden Gebietes ableiten mit dem Ziel, die spezifische Hautveränderung exakt einzugrenzen und herauszuarbeiten.

Eine andere Möglichkeit der Bilderfassung besteht darin, daß ein Laserstrahl geringer Leistung verwendet wird, dessen Energie keinen Einfluß auf das Hautgebiet ausüben kann. Dieser Laserstrahl wird über das zur Behandlung vorgesehene Hautgebiet rasterartig geführt. Das von der Hautoberfläche gestreute Licht wird durch einen optischen Empfänger erfaßt, und die Intensität und/oder der Spektralbereich des gestreuten Laserlichts dient dann zum Aufbau eines Bildes des zu behandelnden Hautgebietes. Auch hierbei kann die Festle-

gung der Behandlungsgrenzen durch Erfassung des Gebietes in verschiedenen Spektralbereichen und anschließenden Bildvergleich bei einzustellender Diskriminatorsschwelle erfolgen.

Eine dritte Möglichkeit besteht darin, das Gebiet mit Hilfe eines Lasers zu erfassen, wobei der Laserstrahl über ein Interferometer, insbesondere Michelson-Interferometer in zwei überlagerte Wellenzüge aufgespalten wird. Mit diesem Austrittsstrahl wird dann die Hautpartie angestrahlt. Topographische Unebenheiten erscheinen dann begrenzt von geschlossenen Hell-Dunkel-Ringen. Das erzeugte Bild wird dann mit einer TV-Kamera aufgenommen, gespeichert und z. B. auf einem Bildschirm dargestellt. Durch vorgegebene Entscheidungsschwellen kann das zu behandelnde Hautgebiet auch auf diese Weise abgegrenzt werden, z. B. zur Behandlung von Warzen, deren Pigmentierung sich üblicherweise nicht von der umgebenden Haut unterscheidet.

Die Erfindung ermöglicht eine nahezu selbsttätige Behandlung von Pigmentierungen der Haut oder topographischen Hautveränderungen. Die exakte Bilderfassung des zu behandelnden Gebietes erlaubt eine Behandlung, die nahezu keine Schädigung angrenzender gesunder Haut hervorruft. Durch eine durch die Raumkoordinaten des zu behandelnden Gebietes bestimmte programmierte Steuerung des Laserstrahls kann eine exakte Dosierung der Auftreffenergie erreicht werden. In einer bevorzugten Weiterentwicklung der Erfindung kann die Energiedichte auch in Abhängigkeit von Raumkoordinaten gesteuert werden, die sich ergeben, wenn eine andere weitere Entscheidungsschwelle für die Erfassung von Raumkoordinaten zugrunde gelegt wird.

Die Erfindung ermöglicht eine exakte Anpassung der Behandlung an Art und Struktur eines zu behandelnden Gebietes. Eine gewünschte Energiemenge kann sehr genau festgelegt werden. Entgegen einem Handbetrieb, bei dem nur die Laserleistung genau einstellbar ist, besteht nach der Erfindung auch die Möglichkeit, die Behandlungsgeschwindigkeit, den Strahldurchmesser und den Zeilenabstand genau festzulegen.

Die Erfindung wird nachstehend anhand eines Ausführungsbeispiels näher erläutert. Es zeigen:

Fig. 1 einen Aufbau einer ersten Ausführungsform der Erfindung,

Fig. 2 einen Aufbau einer zweiten Ausführungsform der Erfindung,

Fig. 3 einen Aufbau einer dritten Ausführungsform der Erfindung,

Fig. 4 ein Blockschaltbild.

Die Wechselwirkung von Licht mit Haut erfolgt über Streuung und Absorption. Die Intensitätsabnahme eines in die Haut eindringenden Lichtstrahls ist annähernd exponentiell. Um eine Schädigung gesunden Gewebes zu verringern, wird für die Zwecke der Erfindung eine Laserwellenlänge gewählt, bei der der Absorptionsweg in gesunder Haut möglichst groß ist. Vorzugsweise liegt die benutzte Laserwellenlänge im Bereich zwischen 400 und 1200 Nanometern. In Frage kommen insbesondere Dauerstrichlaser mit hoher Eindringtiefe in biologisches Material. Vorzugsweise werden Laser mit 5 bis 15 W verwendet. Zur Vereinfachung der Handhabbarkeit und zur Erhöhung der Sicherheit sollten Laser mit sichtbarem Licht verwendet werden.

Das vom Gewebe absorbierte Licht wird in Wärme umgewandelt, die u.a. über Diffusion abgeführt wird. Die Ortsauflösung eines Dermatologielasers wird dementsprechend durch eine die Haut charakterisierende

thermische Diffusionslänge begrenzt. Ein Behandlungserfolg setzt eine Temperaturerhöhung in der Haut von ca. 25°C voraus. Bei 1 cm/Sekunde Behandlungsgeschwindigkeit bedarf es daher einer Laserleistung von etwa 5 Watt.

Der Aufbau der Erfindung ist etwa wie folgt:

Es wird vorzugsweise ein Argonlaser 1 verwendet, der auf einer optischen Bank angeordnet ist. Der Ausgangsstrahl 8 des Lasers 1 ist auf eine Strahlableiteneinheit 5 gerichtet, die derart angeordnet ist, daß ein möglichst optimaler Raumwinkel zur Behandlung des Patienten gewährleistet ist. Als Strahlableiteneinheit wird vorzugsweise ein XYZ-Scanner mit Spiegeln 6 und 7 mit hoher Positioniergenauigkeit benutzt. Zur Steuerung des Scanners werden diesem Anfangs- und Endkoordinaten und die Abtastgeschwindigkeit eines abzufahrenden Vektors übermittelt. Diese Steuerungsdaten werden über eine Rechnerschnittstelle von einem Prozeßrechner geliefert.

Zur Erfassung des zu behandelnden Hautgebietes 2 können in Abhängigkeit von der zu behandelnden Struktur folgende Einrichtungen verwendet werden.

Bei Pigmentierungen, wie z. B. Feuermalen, Tätowierungen, oder ähnlichem kann ein Bilderfassungssystem verwendet werden, das z. B. eine Hell-Dunkel-Unterscheidung ermöglicht. Dazu kann eine Videokamera 4 verwendet werden, die ein Schwarz-Weiß-Bild liefert, das anschließend gespeichert wird. Vorzugsweise wird das zu behandelnde Gebiet durch mehrere spektralbegrenzte Filter z. B. rot, gelb, grün, blau bestrahlt. Die TV-Kamera macht von jeder farbigen Belichtung eine Aufnahme. Zur Festlegung der Grenzen des Behandlungsgebietes werden dann die verschiedenen Aufnahmen durch logischen Vergleich der gespeicherten Bildpunkte miteinander verglichen. Der Bildpunktvergleich kann z. B. additiv oder subtraktiv sein. Im Rahmen der Erfindung liegt auch die Faltung eines Bildes mit einer Faltungsfunktion oder die digitale Filterung, die z. B. auch zweidimensional sein kann. Diese Auswertung erfolgt vorzugsweise in einem Prozeßrechner, der mit einem Optimierungsalgorithmus arbeitet, um das Behandlungsgebiet bestmöglich gegenüber angrenzendem gesunden Gebiet abgrenzen zu können. Die Erfassungsparameter für bestimmte Typen von Hautveränderungen 3 können auch durch den Prozeßrechner gespeichert werden und können dann für spätere Behandlungen die Grundlage der Auswertung bilden. Die Entscheidungsschwelle, die zur Abgrenzung von zu behandelndem von nicht zu behandelndem Hautgebiet gesetzt wird, kann einstellbar gewählt werden, sie kann jedoch auch durch den Prozeßrechner selbst in einem Optimierungsverfahren festgelegt werden.

Eine andere Möglichkeit der Bilderfassung (Fig. 2) besteht darin, daß zunächst ein Laser mit geringer Leistung nach einem fest vorgegebenen Muster rasterartig über die Oberfläche des zu behandelnden Hautgebietes geführt wird. Die Energiedichte dieses Lasers ist derart gering, daß auf diese Weise keine Hautschädigungen hervorgerufen werden. Das von der Hautoberfläche gestreute Laserlicht wird von einem Hohlspiegel 9, der für den Laserstrahl eine Durchgangsöffnung 11 aufweist, und einen optischen Empfänger 10 erfaßt. Die Intensität und/oder der Spektralbereich (z. B. bei Lasern mit durchstimmbarer Wellenlänge) wird unter Zuordnung zu den entsprechenden Auftreffkoordinaten des Laserlichts gespeichert und liegt dann in vergleichbarer Form vor, wie eine Aufnahme mit einer TV-Kamera. Die Auswertung dieser Aufnahme kann auf die gleiche Weise,

wie oben angegeben, vorgenommen werden.

Bei bestimmten Hautveränderungen, die nicht mit einer Pigmentierung verbunden sind, wie z. B. Warzen, ist die Verwendung einer TV-Kamera oder die rastermäßige Erfassung mit Hilfe eines Laserstrahls nicht möglich. In diesem Fall wird der Laserstrahl vorzugsweise über ein in den Laserstrahl geschaltetes Michelson-Interferometer 12 in zwei überlagerte Wellenzüge aufgespalten (Fig. 3), die auf die zu behandelnde Hautpartie gerichtet werden. Unebenheiten erscheinen dann als geschlossene Linien, die mit einer TV-Kamera aufgenommen werden können. Aus einer derartigen Aufnahme können Raumkoordinaten abgeleitet werden, die eine genaue Abgrenzung eines zu behandelnden Gebietes ermöglichen. Zusätzlich läßt sich auf diese Weise der Grad der Erhöhung der Hautveränderung über die gesunde Haut feststellen, da die Zahl der erzeugten Linien einen Grad für die Höhe der Hautveränderung darstellt. Die Höheninformation kann dann z. B. zur Steuerung der Energiedichte des Behandlungslasers auf dem zu behandelnden Hautgebiet verwendet werden. Vorgenannte Maßnahmen können auch kombiniert werden, so daß jeweils eine optimale Erfassung in Abhängigkeit von dem Typ der Veränderung erfolgen kann.

Die erfindungsgemäße Einrichtung ermöglicht die Festlegung der Abtastgeschwindigkeit, der Laserleistung, des Spotdurchmessers und des Zeilenabstandes des rasterartig über das zu behandelnde Hautgebiet geführten Behandlungslasers. Die über das Bilderfassungssystem 15 gewonnenen Raumkoordinaten werden gemäß Fig. 4 in einer Rechneinheit 14 zu einer Befehlsfolge umgewandelt, die auf den XYZ-Scanner (5) gegeben wird, der den Lichtstrahl des Lasers exakt positioniert. Die Wahl von Abtastgeschwindigkeit, Laserleistung, Spotdurchmesser und Zeilenabstand kann selbsttätig durch die Rechneinheit bei Vorgabe des Typs der Hautveränderung an der Bedieneinheit 13 festgelegt werden.

Damit die durch das Bilderfassungssystem festgelegten Raumkoordinaten sich während der Behandlung mit dem Laserstrahl nicht verändern, muß das zu behandelnde Hautgebiet exakt in bezug auf die Behandlungseinrichtung positioniert werden. Bei räumlichen Stellungsänderungen des zu behandelnden Hautgebietes ist eine erneute Erfassung der Raumkoordinaten erforderlich. Zur Vermeidung dieses Nachteils kann vorgesehen sein, daß die absoluten Koordinaten des zu behandelnden Gebietes z. B. über zwei gekennzeichnete Fixpunkte erfaßt werden, daß deren räumliche Lage ständig während der Behandlung kontrolliert wird und bei Veränderungen eine Koordinatentransformation der Behandlungsraumkoordinaten bewirkt, so daß bei einer Bewegung des Patienten während der Behandlung eine Nachsteuerung des Laserstrahls erfolgen kann.

Bezugszeichenliste

- 1 Laser
- 2 Gebiet
- 3 Hautveränderung
- 4 Kamera
- 5 X-Y-Z-Scanner
- 6 Spiegel
- 7 Spiegel
- 8 Laserstrahl
- 9 Hohlspiegel
- 10 Reflektor
- 11 Öffnung

- 12 Interferometer
- 13 Bedieneinheit
- 14 Prozeßrechner
- 15 Bilderfasser

Patentansprüche

5

1. Einrichtung zur Behandlung von Hautveränderungen des menschlichen oder tierischen Körpers mit Hilfe eines Laserstrahls (8), der über das zu behandelnde Hautgebiet (2) führbar ist, **dadurch gekennzeichnet**, daß ein Bilderfassungssystem (4) zur Aufnahme und Speicherung der Bilder eines zu behandelnden Hautgebietes (2) vorgesehen ist, daß durch einen Diskriminator aus dem gespeicherten Bild Raumkoordinaten für das zu behandelnde Gebiet abgeleitet werden, und daß aus den Raumkoordinaten durch eine Steuereinrichtung eine Befehlsfolge abgeleitet wird, die zur Steuerung der Auftreffkoordinaten des Laserstrahls auf dem zu behandelndem Gebiet verwendbar ist.

2. Einrichtung nach Anspruch 1, dadurch gekennzeichnet, daß das Bilderfassungssystem eine Videokamera (4) ist, mit deren Hilfe mehrere Aufnahmen des Behandlungsgebietes erfolgen, daß eine Auswertung der gespeicherten Bilder in verschiedenen Spektralbereichen erfolgt, wobei durch einen Bildvergleich der gespeicherten Aufnahmen bei wählbarer Diskriminatorschwelle die Raumkoordinaten des zu behandelnden Gebietes abgeleitet werden.

3. Einrichtung nach Anspruch 1, dadurch gekennzeichnet, daß das Bilderfassungssystem einen Laser geringer Leistung aufweist, dessen Strahl mit festen Anfangs- und Endkoordinaten rastermäßig über das zu behandelnde Hautgebiet geführt wird, daß das von der Hautoberfläche gestreute Licht durch einen optischen Empfänger (10) erfaßt und die Intensität und/oder der Spektralbereich des gestreuten Laserstrahls unter Zuordnung zu den entsprechenden Auftreffkoordinaten des Laserlichts gespeichert wird, und daß ein derart gespeichertes Bild mit weiteren Bildern des zu behandelnden Hautgebietes bei wählbarer Diskriminatorschwelle verglichen wird und daraus die Raumkoordinaten des zu behandelnden Gebietes abgeleitet werden.

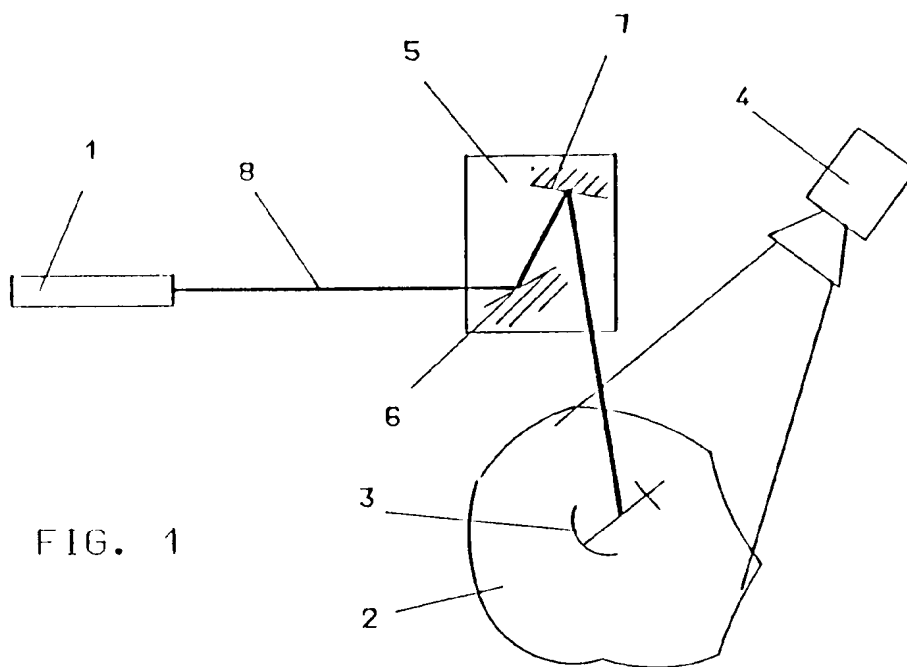
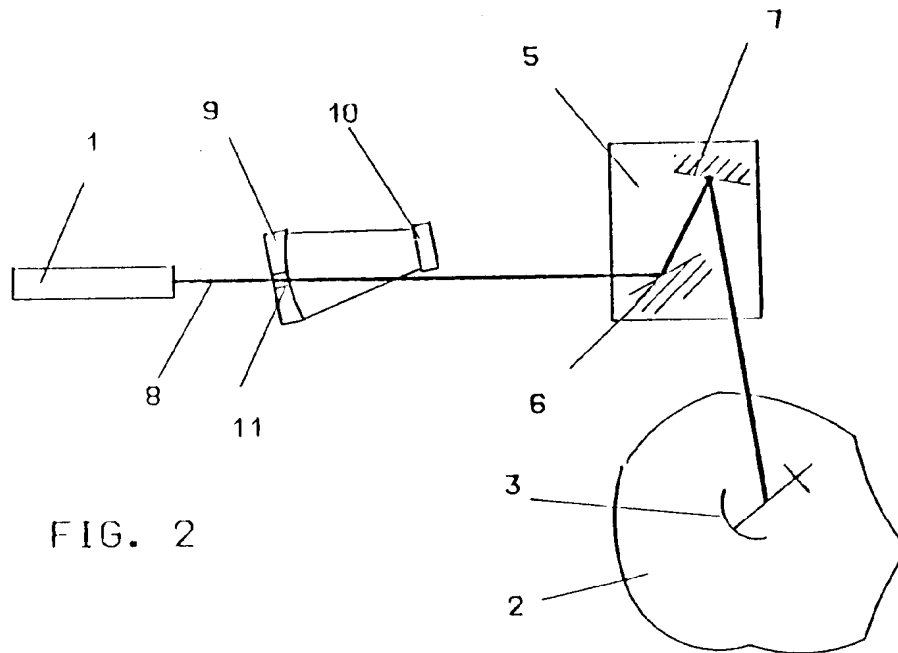
4. Einrichtung nach Anspruch 2 oder 3, dadurch gekennzeichnet, daß einzelne Bildpunkte oder Bildpunktgruppen der gespeicherten Bilder logisch miteinander verknüpft werden und aus einem derartig verknüpften Bild die Raumkoordinaten des zu behandelnden Gebietes abgeleitet werden.

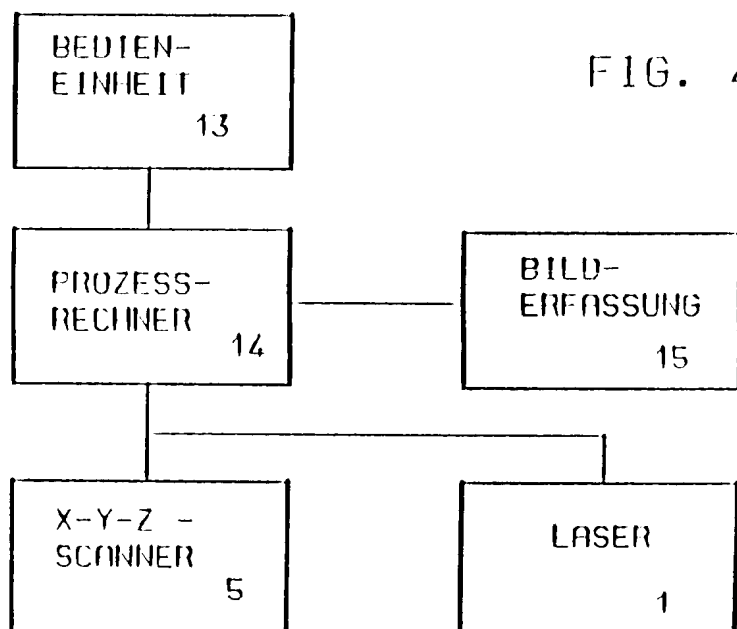
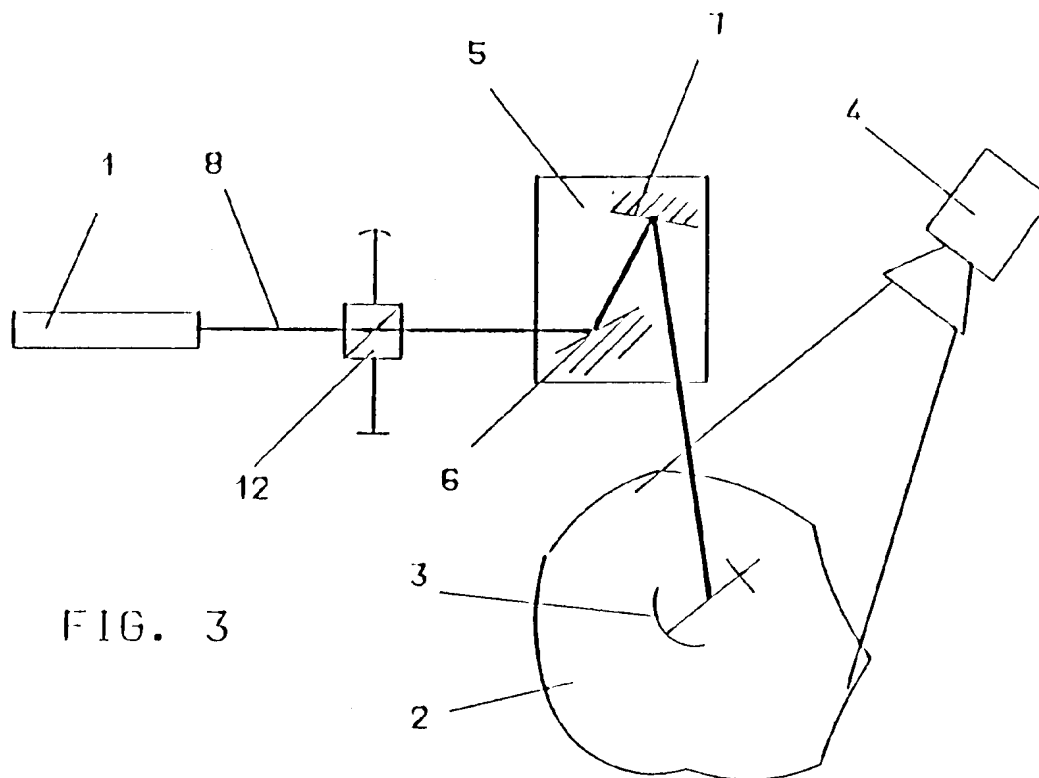
5. Einrichtung nach Anspruch 1, 2 oder 3, dadurch gekennzeichnet, daß das Bilderfassungssystem einen Laser geringer Leistung aufweist, dessen Strahl nach Durchlaufen eines Interferometers (12) auf das zu behandelnde Hautgebiet gerichtet wird, und daß aus dem mittels einer Kamera (4) erfaßten und gespeicherten Bildes des angestrahlten Hautgebietes dessen Raumkoordinaten bei Zugrundelegung einer wählbaren Diskriminatorschwelle ermittelt werden.

6. Einrichtung nach Anspruch 5, dadurch gekennzeichnet, daß das Interferometer (12) ein Michelson-Interferometer ist.

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Hierzu 2 Seite(n) Zeichnungen





(12)

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(54) **Method and apparatus for photoepilation.**

(57) The certain quantity of light energy sufficient to effect lifelessness in each of a particular type of hairs is provided to the tip of a hand held probe (18) each time a switch (22) is closed, independent of the duration of the switch closure. A foot switch (22) is actuated to provide a series of timed and spaced pulses of light energy for a period selectable by rotating a wiper of a variable resistor-capacitor circuit. The sum of

the energies of the pulses over the selected period equals the certain quantity of light energy. Alternatively, four foot switches can be provided, one of which can be depressed to provide one of four fixed periods of light pulses. Commonly, the certain quantity of light energy is first determined for the hair type and then that certain quantity is used for all the hairs of that type to be removed.

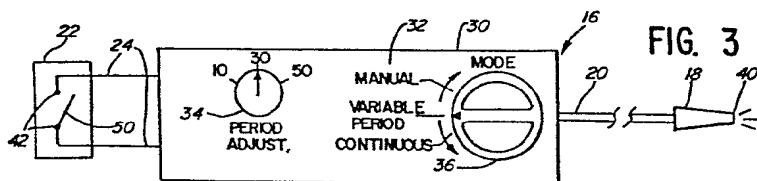


FIG. 3

- 1 -

This invention relates to epilation primarily for cosmetic and medical purposes, and specifically relates to an improved method of performing photoepilation and an apparatus for performing the method.

Photoepilation is the removal of hair using light energy and is a known, commercially available process. It is performed to obtain cosmetically more pleasing skin by removal of unsightly hair from locations such as the face, legs, arms and back.

It is performed by an electrologist or operator seated opposite a patient and specifically opposite an exposed area of the patients skin on which the epilation is to be performed. The operator uses a magnification means such as wearing extended loupe eye glasses to sight on the hair follicles, and in one hand holds a light probe while in the other hand holds a tweezers, for removal of the hairs from the follicles after application of the light probe. The light probe is the termination of a flexible optical means, such as a bundle of optical fibers, that carry light energy to the hair follicle from a discharge lamp contained in a housing of a photoepilation device. The device includes the probe, the housing and a foot switch, which the operator depresses to control the number of light pulses to be provided by the device.

Performing the photoepilation is demanding of the operator's mental and physical faculties. The operator focuses his or her eyes on one hair and its follicle by looking through the extended loupe eye glasses, moves his or her head to the proper distance from the hair to obtain a sufficient magnification and/or clear focus and then holds his or her head still to maintain the desired sight of the follicle.

- 2 -

The operator then moves the tip of the probe, which sources the light energy, to the hair follicle and positions it for applying properly the light energy to the hair follicle. The operator then depresses his or her foot on the foot pedal to close the switch therein and energize the photoepilation device, providing the light energy in timed and spaced pulses. The number of pulses, or the time (which is proportional to the number of pulses) is counted by the operator either out loud or privately to apply the proper quantity of light energy to the hair follicle to effect lifelessness therein. When the proper quantity of energy has been applied to the hair to kill the hair body or root, the operator lifts his or her foot from the foot pedal to stop production of the light energy, and removes the hair from the follicle with the tweezers held in his or her other hand. The operator then moves to the next hair to be removed and repeats this entire demanding procedure.

The quantity of light energy produced by the device is regulated or dependant entirely and solely upon the length of time that the operator depresses the foot pedal of the switch assembly.

The electrologist, thus, to remove one hair, must have excellent motor skills to coordinate simultaneous movement of his or her head and eyes, both hands and one foot. Further, this procedure is tiring because of the physical and mental demands placed upon the electrologist to perform for extended periods.

It is desirable to alleviate at least one of the demands made of the electrologist.

Accordingly, the invention provides a method of photo-epilation of a plurality of hair bodies of at least
5 one type wherein light energy is conducted from a source to the hair root or the vicinity thereof by way of the tip of a probe by actuating a switch coupled to the source for the duration required to provide light energy sufficient to effect lifelessness of the hair body and
10 repeating such process for affecting each of the plurality of hair bodies, characterized by the steps of selecting a certain period and applying the light energy by actuating the switch for a duration independent of said period selected.

15 Further, the invention provides an apparatus for performing photoepilation of a plurality of hair bodies of at least one type, including a source of light energy, a probe terminating at a tip and an optical cable coupled
20 between the source and the tip for conducting light energy therebetween and a switch for energizing said source; characterized by a control circuit coupled between the source and the switch and capable of being energized by actuation of the
25 switch to produce light energy for a certain period independent of the length of time the switch is actuated.

The preferred embodiments of this invention now will be
30 described, by way of example, with reference to the drawings accompanying this specification in which:

Figure 1 depicts a scene in which an electro-
logist is performing a photoepilation
35 procedure upon the hairs on the back of a patient;

Figure 2 is a block diagram of the one embodiment of the apparatus of the invention;

Figure 3 is a front side view of another embodiment of the apparatus of the invention; and
5 Figure 4 is a block diagram of the other embodiment of the apparatus of the invention.

10 In Figure 1, an electrologist or operator 10 is removing hair from the back 12 of a patient 14 using a proximately located photoepilation apparatus indicated generally by the reference character 16. Apparatus 16
15 comprises a probe 18 held in the right hand of electrologist 10 and an optical cable 20 connecting the probe 18 to the apparatus 16. Optical cable 20 comprises flexible optical transmission means such as optical
20 fibers. Apparatus 16 further includes a foot switch assembly 22 connected thereto by an electrical cable 24. Assembly 16 is operatively located under the right foot of the electrologist. The electrologist 10 comfortably is seated opposite the bare back 12 of the seated patient 14. The electrologist 10 wears an extended loupe carrying eyeglasses 26 better to see the hair
25 follicles on the patient's back 12 and holds a tweezers 28 in his or her left hand for removing lifeless hairs.

Typically all of the hairs or hair bodies to be removed from one region of a person's skin, such as face,
30 arms, back, legs, are of one to three types, and the light energy that must be applied to each hair of one type to effect lifelessness therein is constant. Thus, once a particular quantity of light energy to effect lifelessness is established for a hair type, that
35 quantity need only be repeatedly supplied to each hair root to perform the photoepilation.

In Figures 3 and 4, apparatus 16 includes a box-like housing or container 30 providing a face plate 32 on which are located a PERIOD ADJUST knob 34 and a MODE CONTROL knob 36. Interior of container 30, a pulsed light source 38, including a flash lamp, produces or sources light energy through the optical cable 20 to probe 18. The light energy is output from the tip 40 of the probe. The foot switch assembly 22, having one pair of normally open contacts 42 is connected to the apparatus by electrical cable 24. A control circuit 41, comprising mode selector circuit 44, timer circuit 46 and variable RC circuit 48, operate generally to produce timed and spaced pulses of light energy in source 38, and thereby to probe tip 40. Mode selector circuit 44 determines whether the flash lamp will be operated in one of three modes:

manually, for a period of time commencing and ending respectively with the closing and opening of contacts 42 by wiper or foot pedal 50; for a variable period commencing with the closing of contacts 42 by foot pedal 50 and extending for a duration determined by the setting of the variable RC circuit by knob 34 through shaft 52; and continuously for a period of time commencing with the closing of contacts 42 by foot pedal 50 and ending with a second closing of contacts 42 by foot pedal 50. In all cases, contacts 42 are closed by electrologist 10 depressing his or her foot on foot pedal 50 and are opened by the technologist raising his or her foot from the foot pedal. Rotation of knob 36 through shaft 54 effects selection of the desired mode.

In the manual and continuous modes, closing of contacts 42 is conducted to mode control circuit 44 by wires 56 from cable 24. In the variable period mode,

cable 24 directly carries the signal indicating closing of contacts 42 to variable RC circuit 48 and timer circuit 46. Timer 46 then produces a signal on wire 58 to selector 44 that has a period corresponding to that indicated by knob 34. In all three cases, mode selector circuit 44 includes a circuit to energize light source 38 and obtain regular timed and spaced pulses of light therefrom. The timing and spacing of the pulses is fixed to obtain equal but incremental quantities from each pulse of light energy. The only variable afforded by the control circuit 41 thus is the number of pulses or incremental quantities of light energy sourced to probe tip 40. The control circuit 41 does not count the number of pulses but provides them for a selected period.

In operation, the electrologist selects the variable period mode by rotating knob 36 and selects a certain period corresponding to the quantity of light energy, in the form of the timed and spaced pulses, required to be sourced at the tip 40 to effect lifelessness in a particular type of hair body. Selection occurs by rotation of knob 34 to vary the resistance or capacitance of circuit 48. The electrologist then adjusts his or her head to sight on one hair follicle, properly places the probe tip 40 in the vicinity of the follicle and depresses the foot pedal wiper 50 once. The control circuit then automatically, and independently of the length of time that the contacts 42 are closed, causes a series of timed and spaced pulses of light energy, for the selected period, to be sourced to the probe tip 40, and therefrom to the hair body to effect lifelessness therein. The electrologist then removes the lifeless hair body from the follicle, and moves to the next follicle, repeating the described procedure.

The method and apparatus of the invention thus relieve the electrologist from the manual counting of time or pulses to effect the lifelessness with the light energy. The termination of the selected period is seen by the electrologist by the lack of light pulses being supplied to the follicle from the probe tip. Operation in the manual and continuous mode is similar to that described in the previous paragraph but requiring a different actuation of the foot pedal for each respective mode.

In Figure 2, a second embodiment of the photoepilation apparatus of the invention is indicated generally by reference character 60. The same reference characters indicate the same elements in Figures 2, 3 and 4.

Apparatus 60 comprises container 30, optical cable 20, probe 18, probe tip 40 and pulsed light source 38 previously described. A control circuit 62 includes a timer circuit 64, four resistor-capacitor networks RC1-RC4, respectively, 66-72, a switch assembly 74 and an electrical cable 76. Assembly 74 provides four, normally open pairs of contacts or switches 78-84. Switches 78-84 are mounted on a panel for operation by the electrologists's foot and may be such as push button switches. Four pairs of wires 86 connect the contacts of each switch to its respective resistor-capacitor circuit, these wire pairs forming cable 76. In turn, each resistor-capacitor circuit 66-72 is connected to timer circuit 64 by four separate wires 88. Timer circuit 64 is connected to light source 38 by wire 90.

Control circuit 62 provides four fixed and individually selectable periods of production of the light energy pulses from light source 38 corresponding to the four

switches 78-84. Switch 84 provides a period of fifteen (15) minutes, switch 82 provides a period of ten (10) seconds, switch 80 provides a period of thirty (30) seconds and switch 50 provides a period of fifty (50) seconds.

Actuating any one switch 78-84 energizes the corresponding resistor-capacitor circuit and causes timer circuit 64 to produce pulsing signals on wire 90 for the selected period. The pulsing signals on wire 90 cause the flash lamp in light source 38 to flash at regular timed and spaced intervals for the period, providing equal energy light energy pulses at probe tip 40.

The photoepilation procedure performance with apparatus 60 is similar to the procedure performance with apparatus 16, except there is no setting of a mode or period. Instead, selecting a certain period occurs by selecting the desired foot actuated push button switch, which then is depressed to source the required quantity of light energy at the probe tip. Sourcing the light energy pulses is independent of the length of time that the selected switch is actuated.

Selection of the period or energy necessary to effect lifelessness in a particular type of hair is by experiment or otherwise as desired.

Known timing circuits, including monostable multivibrators, can provide the period signals in an apparatus 60. One circuit may provide each fixed period or one circuit may provide all four fixed periods. A known timing circuit, such as a monostable multivibrator can provide the single variable period circuit in apparatus 16. In either case, a circuit such as a monostable

multivibrator or one shot, provides a timing signal having a period independent of the switch closure or
5 actuation time. This frees the electrologist from the requirement of having to regulate or control, by counting, the light energy applied to each hair body.

Different control and timing circuits, and switch
10 arrangements and locations can be employed.

Thus, the invention involves providing a certain quantity of light energy at the tip of a probe sufficient to effect lifelessness in a hair body upon the closure of
15 a pair of switch contacts, independent of the duration of the closure of said switch contacts. The quantity of light energy is provided in the form of incremental flashes or pulses of light energy, each of which having a light energy quantity equal to the others, but less
20 than the proper quantity. The pulses are provided in like, timed and spaced relation to one another so that providing a series of the pulses for a selected duration period provides the proper quantity of light energy. In effect, the sum of the energies of the light pulses
25 provided in the selected period equals the proper quantity of light energy to effect the lifelessness.

The demands upon the electrologist are alleviated by the electrologist selecting the period by setting the
30 period duration to obtain the proper quantity of light energy to effect the lifelessness in a particular type of hair located on such as a face, an arm or a back. Then for each hair of the type, the electrologist need only actuate a switch once to obtain the proper quantity of light energy. The duration of the actuation of
35 the switch need not be precisely regulated by the

electrologist because the duration of the selected
period is independent of the duration of switch clo-
5 sure.

The apparatus of the invention comprises a flasch
lamp and control curcuit assembly, a hand held probe
connected to the flash lamp by a bundle of flexible,
10 optical fibers and a foot switch assembly electrically
connected to the control circuit by a cable. The con-
trol circuit is operable, in reaction to actuation of
the foot switch, to flash the lamp in timed and spaced
pulses of equal light energy.

15 In one embodiment, the foot switch assembly comprises
a plurality of switches, one switch for each of a plu-
rality of fixed periods. Selecting the certain period
then comprises placing the operator's foot above the
20 switch corresponding to the quantity of light energy
required to effect the lifelessness in the type of
hair to be removed from the subject patient. Alterna-
tively, the control circuit includes a variable timer
for selecting the certain period and the foot switch
25 assembly comprises one switch for commencing the pe-
riod. Selecting the period thus comprises moving the
wiper of a variable resistor or capacitor to a parti-
cular setting.

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CLAIMS:

1. A method of photoepilation of a plurality of hair bodies of at least one type wherein light energy is
5 conducted from a source to the hair root or the vicinity thereof by way of the tip of a probe by actuating a switch coupled to the source for the duration required to provide light energy sufficient to effect lifelessness of the hair body and repeating such process
10 for affecting each of the plurality of hair bodies, characterized by the steps of selecting a certain period and applying the light energy by actuating the switch for a duration independent of said period selected.
15
2. The photoepilation method according to claim 1 characterized in that the step of selecting a certain period includes varying the components of a timer circuit.
20
3. The photoepilation method according to claim 1 characterized in that the step of selecting a certain period includes selecting one of a plurality of switches, each switch being associated with one of a
25 plurality of certain periods.
4. An apparatus for performing photoepilation of a plurality of hair bodies of at least one type, including a source of light energy, a probe terminating at
30 a tip and an optical cable coupled between the source and the tip for conducting light energy therebetween and a switch for energizing said source; characterized by a control circuit (41, 62) coupled between the source (38) and the switch (22, 74) and capable of being
35 energized by actuation of the switch to produce light energy for a certain period independent of the length of time the switch is actuated.

5. The photoepilation apparatus according to claim 4, characterized in that said control circuit includes at
5 least one timer circuit (46, 64) operating for said certain period to produce said light energy independent of the length of time the switch (22, 74) is actuated.

6. The photoepilation apparatus according to claim 5,
10 characterized in that said timer circuit (46) includes variable components (48) for setting said certain period.

7. The photoepilation apparatus according to claim 5,
15 characterized in that said timer circuit (64) includes a plurality of fixed components (66, 68, 70, 72) arranged to provide a plurality of certain periods and there are a plurality of switches (78, 80, 82, 84) one for each certain period, actuation of one switch energizing
20 the control circuit (62) for one certain period.

8. The photoepilation apparatus according to any one of claims 4 to 7, characterized in that all switches are foot-actuated switches.

25
9. The photoepilation apparatus according to any one of claims 4 to 8, characterized in that the probe is portable and capable of being hand-held during operation of said apparatus.

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FIG. 1

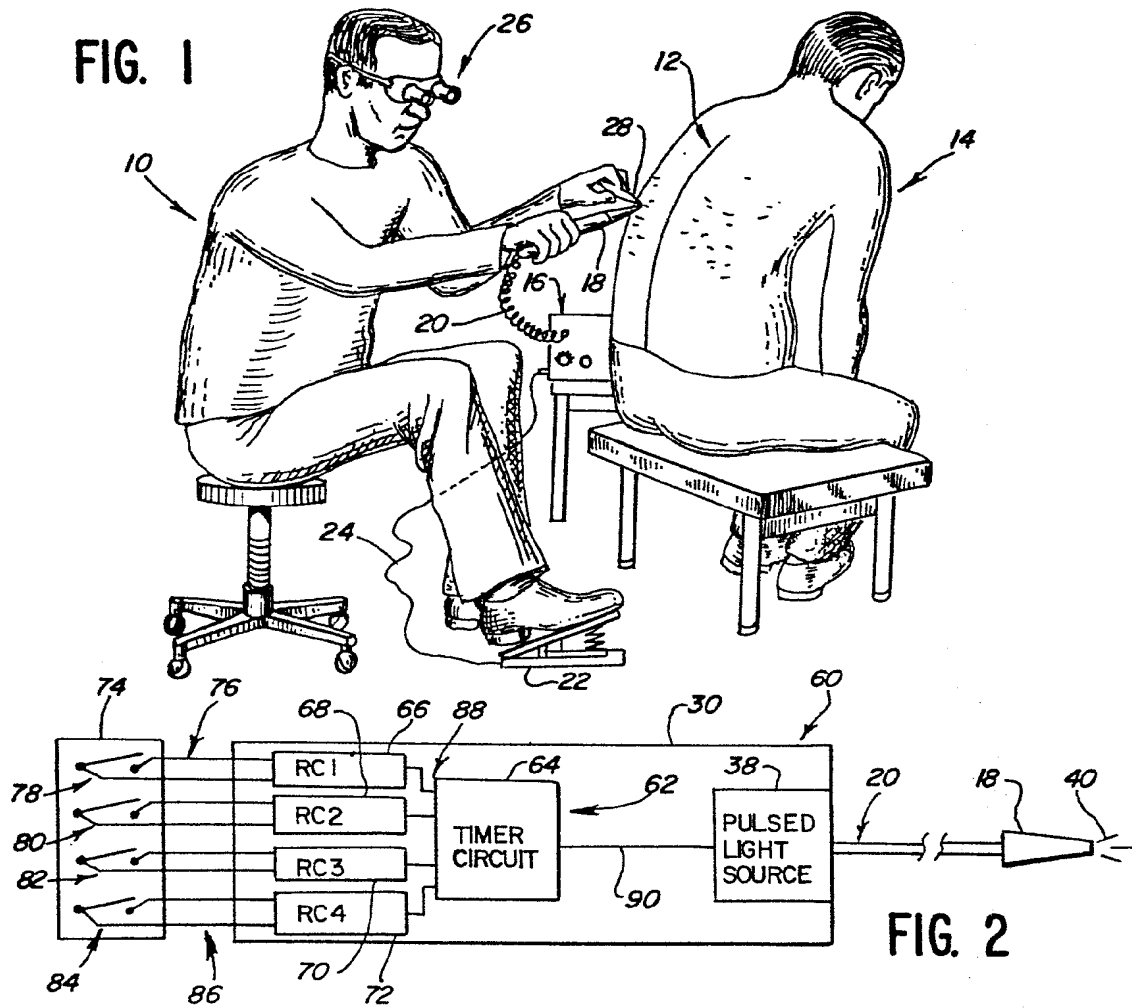


FIG. 2

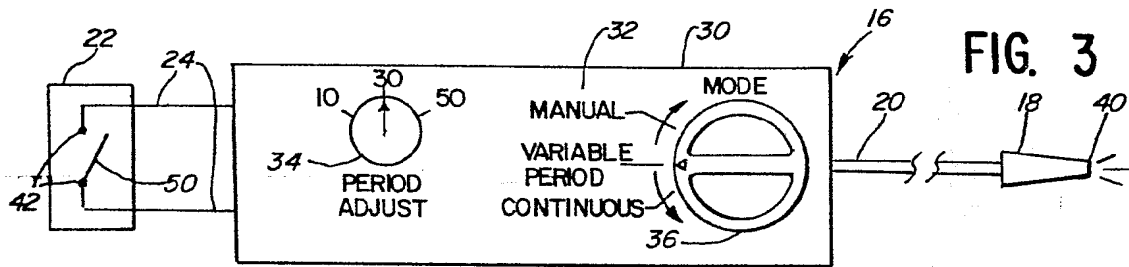


FIG. 3

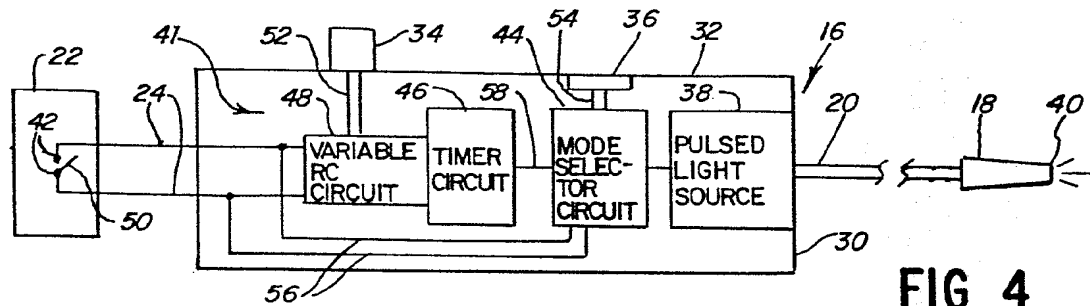
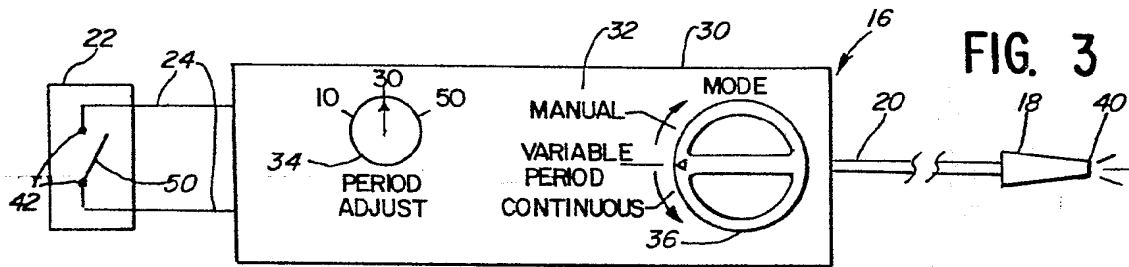


FIG. 4



European Patent
Office

EUROPEAN SEARCH REPORT

0142671
Application number

EP 84 11 1249

DOCUMENTS CONSIDERED TO BE RELEVANT			
Category	Citation of document with indication, where appropriate, of relevant passages	Relevant to claim	CLASSIFICATION OF THE APPLICATION (Int. Cl.4)
Y	US-A-4 388 924 (WEISSMAN et al.) * whole document *	1-9	A 61 B 17/36
Y	US-A-3 693 623 (HARTE et al.) * figures; column 3, lines 6-22; claims 1-7 * -----	1-9	
			TECHNICAL FIELDS SEARCHED (Int. Cl.4)
			A 61 B A 61 F
The present search report has been drawn up for all claims			
Place of search THE HAGUE		Date of completion of the search 20-12-1984	Examiner LOWE D.
CATEGORY OF CITED DOCUMENTS			
X : particularly relevant if taken alone Y : particularly relevant if combined with another document of the same category A : technological background O : non-written disclosure P : intermediate document		T : theory or principle underlying the invention E : earlier patent document, but published on, or after the filing date D : document cited in the application L : document cited for other reasons & : member of the same patent family, corresponding document	

12 **EUROPEAN PATENT APPLICATION**

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51 Int. Cl. 4: **A 61 B 17/36**
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54 **Laser system for providing target specific energy deposition and damage.**

57 A hand piece (20) for use with a laser (14) includes a scanning mechanism which controls dosimetry of radiation applied to a target area (12) which is adjustable to limit thermal diffusion from the light absorbing portion of the irradiated target site for selective target specific energy deposition. When used for dermatologic purposes, the adjustable scanning mechanism permits radiation to impinge on tissue for a predetermined period of time for the selective necrosis of highly-filled blood vessels, while leaving adjacent tissue and empty blood vessels undamaged. The dwell time of the laser beam is designed to match the diffusion time for thermal destruction of the wall of the abnormal vessel, with the dwell time adjusted by the scanning rate.

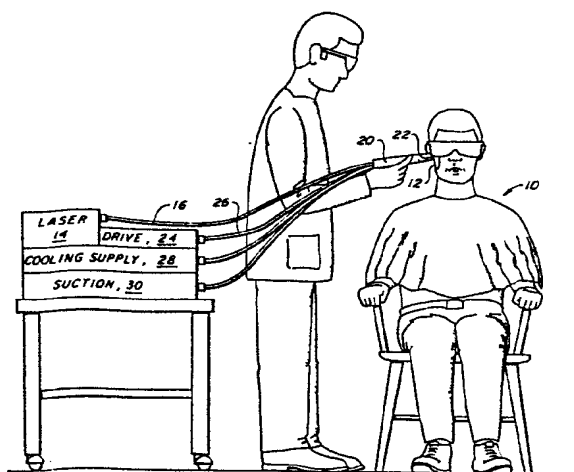


FIG. 1

FIELD OF INVENTION

This invention relates to medical treatment and, more particularly, to the control of a laser output in such a way as to limit the dwell time on any particular portion of the tissue, thus permitting target specific energy deposition and damage which is selective and controllable.

BACKGROUND OF THE INVENTION

There are many medical conditions, the treatment of which is substantially improved by being able to control the deposition of laser energy in a specific target tissue in order to damage that target tissue while sparing the adjacent tissue. While those in the past have utilized lasers, particularly in the port wine stain (PWS) syndrome, to destroy blood vessels, the problem associated with such systems is that the dwell time of the laser over the target produces significant thermal diffusion which damages not only the abnormal PWS vessels which are ectatic, i.e., dilated and filled, and strongly absorb the radiation, the ones producing the wine stain, but also damages a significant depth of the dermis such that scabbing and sloughing occurs as a consequence of treatment. Additionally, the use of an anesthetic is prescribed because of the amount of energy imparted to the target area which is painful to the patient. It will be appreciated that the desired treatment for port wine stains is to necrose only those vessels producing the stain while leaving most of the surrounding collagen and the normal vessels undamaged. This translates into control of thermal diffusion, which up until the present time has been difficult either because of the relatively long pulse lengths of the shuttered CW lasers utilized or because of the relatively short pulse lengths of the dye lasers which do not completely destroy the abnormal vessels.

1 Consequently, no successful control has heretofore been exercised to limit the
volume affected by thermal diffusion. The result of the lack of control is that
lasers which dwell on a given target area for 20 milliseconds or more produce so
5 much thermal diffusion that scabbing and sloughing of the epidermis and portions
of dermis are produced regardless of wavelength, assuming any kind of therapeutic
levels are introduced to the target area.

10 By way of further background, if it is desirable to destroy abnormal tissue
contained within the volume of normal tissue and spare the overlying normal layer,
differential absorption of the light is required. This can be obtained by either
intrinsic optical qualities of the target or tagging with some exogenous chromo-
15 phore. In the latter case, the wavelength or color of the laser must be selected on
the basis of the absorption spectra of both the target and the surrounding tissue. In
other words, it should be a wavelength where the target tissue is a good absorber
and the surrounding tissue is a poor absorber. Moreover, the irradiation time must
20 be selected on the basis of the thermal properties of the target and surrounding
tissue and the geometric shape and dimensions of the tissue structure.

Thus, in some cases the target tissue is distinguished from surrounding tissue
by a difference in absorption spectra either due to a naturally-occurring chromo-
25 phore (absorbing molecule) or due to the selective deposition of some dye used in
the treatment regimen. One important example of such a target tissue present
throughout the body is the vasculature which contains erythrocytes. The erythro-
30 cytes contain hemoglobin, a naturally-occurring chromophore with a broad usable
absorption band in the visible. The entire range of visible wavelengths shorter than
approximately 600 nm (nanometers) and extending into the ultraviolet is available
to purposely inflict damage to target tissues containing this chromophore. The
35 specific wavelength selected depends on the relative effects of scattering, which
varies with wavelength; the presence of other chromophores, such as melanin, in
the adjacent or overlying tissues; and the availability of light sources.

In selecting the exposure time, one is limited by the time which will confine thermal damage due to heat transport to an acceptable distance from the target. For the treatment of port wine stains, it is desirable to be in the regime where the dominant mechanism for heat transport is conduction. A characteristic thermal diffusion length for heat conduction is given by

$$L^2 = 4Kt$$

where L = distance that heat diffuses; K = thermal diffusivity coefficient; and t = time allowed for diffusion.

This formula varies slightly with the geometry of the irradiated target in surrounding media, but the variations are not significant. A typical thermal diffusivity coefficient for biological tissues is $0.0015 \text{ cm}^2/\text{second}$.

Considering, for example, the treatment of port wine stains, a type of hemangioma that consists of hypertrophic capillaries in the dermis causing a pink, red or purple coloring of the skin, the pink and red lesions are high in erythrocytes carrying oxygenated hemoglobin (HbO_2), while the purple lesions contain large quantities of deoxygenated hemoglobin (Hb). The lesions are characterized as consisting of abnormal capillary structure with the capillaries varying in diameter, the mean diameter being about 50 micrometers. The wall of the vessel, however, is only a few micrometers thick. The average vessel spacing is 100 micrometers. In order to damage the vessel containing the target hemoglobin, its wall, and a small portion of collagen surrounding the wall, the latter two being relatively free of chromophore, it is important to select an exposure time corresponding to thermal diffusion over a characteristic length slightly greater than the wall thickness. This, in general, refers to the delivery of radiation to a given target area of less than one millisecond. Those prior art devices which deliver 20 milliseconds or more cause damage due to thermal diffusion of heat to a distance

1 even greater than the vessel spacing. Thus the entire tissue bulk is heated by the
vessel network embedded within it and the damage is not at all selective.

J. L. Finley, S. H. Barsky and D. E. Geer in an article entitled "Healing of
5 Port Wine Stains After Argon Laser Therapy," Archives of Dermatology, 1981,
Volume 117, pps. 486-489, and J. L. Ratz, P. L. Baillin, and H. L. Levine in an
article entitled " CO_2 Laser Treatment of Port-Wine Stains: A Preliminary
Report," J. Dermatol. Surg. Oncol. 1982, vol. 8, no. 12, pps 1039-1044, describe
10 both argon lasers and carbon dioxide lasers used in the clinical treatment of port
wine stains. Neither of these provide an optimal treatment modality as the dwell
time is not limited to prevent thermal diffusion.

15 It will be appreciated that the CO_2 laser radiation whose wavelength is
approximately 10 micrometers is very strongly absorbed in water and most proteins.
In port wine stains, both the abnormal vasculature and the surrounding dermal
tissues are approximately 90% water and consequently absorb the incident laser
20 radiation and are heated to the point of thermal necrosis. Thus this treatment does
not involve any specificity of damage. The necrotic tissue eventually sloughs off
and is replaced via the normal healing process by scar tissue formation. Since the
scar tissue formed is usually flat and white, it is often more acceptable to the
25 patient than the original dark and, sometimes, hypertrophic lesion.

Present treatment of port wine stains with the argon laser is performed
using comparatively long pulse times. Because of this, heat has time to diffuse to
30 the surrounding tissue, and the effect observed is the same as for the CO_2 laser in
which radiation is uniformly absorbed in both vascular and surrounding tissue. The
similarity of clinical results with the CO_2 and the long pulse argon lasers has been
noticed and documented in an article by J. W. Buecker, J. L. Ratz and D. F.
35 Richfield entitled "Histology of Port Wine Stains Treated with CO_2 Laser," Fifth
International Conference of Laser Medicine and Surgery, Detroit 1983, in an

abstract. Thus the nonspecificity of prior art laser treatment of port wine stains is both documented and explainable by the relatively long irradiation times causing massive long-distance thermal diffusion for argon lasers and the non-specific absorption for CO₂ lasers.

By contrast ultrashort laser pulses have been used. Studies by R. R. Anderson and J. A. Parish entitled "Microvasculature Can Be Selectively Damaged Using Dye Lasers: A Basic Theory in Experimental Evidence in Human Skin," Lasers in Surgery and Medicine, 1981, Volume 1, pps. 263-276, show that when utilizing pulse dye lasers with fluence level on the order of 3 to 5 Joules/cm² and exposure times of approximately 300 nanoseconds (ns) target specific damage may be produced in normal blood vessels. The wavelength used was 577 nanometers. While the above pulse width was short enough to restrict thermal diffusion to a small portion of the individual erythrocytes having typical dimensions of 7-15 micrometers carrying the hemoglobin, and thermal diffusion subsequent to the pulse could have allowed heating of the containing vessel without damage to the surrounding tissues, the short 300-nanosecond duration caused the vessels to burst and to spew forth blood. It is possible that a shock wave produced by the ultrashort pulse ruptured the blood vessels causing formation of purpura. Since the vessels are 50μ in diameter and the wall is about 1 μ thick, the pulse is so short that only the hemoglobin itself (which is the optical absorber) and any spot on the inner edge of the wall which happens to be in intimate thermal contact with the hemoglobin bearing portion of an erythrocyte are heated during the pulse. After the pulse, the peak temperature achieved within the vessel decays as heat diffuses away. While regions outside the vessel are in fact heated as this diffusion occurs, it is not possible to achieve thermal damage to an adequate depth to insure permanent vessel necrosis.

1 Note that at the present time several microseconds is the longest pulse time
available from commercial pulsed dye lasers. This is still too short to achieve the
desired effect.

5 In summary, it will be appreciated that the difficulty in the prior art
methods of utilizing argon laser treatment lies not in the wavelength, at least for
low melanin skins, but in the exposure time utilized. The argon lasers are CW
lasers which are mechanically shuttered to provide pulsewidths which may vary
10 from 20 milliseconds to 100 milliseconds or more. Even the shortest of these
exposure times, 20 milliseconds, results in thermal diffusion to a length of 100
micrometers which is equal to the average spacing between targets. Thus, even if
15 the laser power is initially absorbed only in the target volumes, thermal diffusion
during the laser pulse itself provides nearly uniform heating of the entire irradiated
area. In order to achieve true specificity, damaging only the target vessels, the
exposure time must be limited to about one millisecond or less, which is too short
20 to be achieved with the mechanical shutters presently in use. Additionally,
nanosecond pulses from dye lasers cause blood vessel rupture and causes only
partial necrosis. Thus these techniques are not optimally useful in treating port
wine stains.

25 Note the following U.S. patents deal with scanning lasers: 3,362,007;
3,642,007; 4,069,823; and 4,316,467; whereas U.S. patents dealing with coaxial bi-
laser beams include 3,456,651; 3,710,798; 3,769,963; 3,906,953; 3,910,276; 4,240,431
30 and 4,408,602. Finally, U.S. patent 3,434,476 deals with a plasma arc scalpel.

SUMMARY OF THE INVENTION

35 In order to limit thermal diffusion, apparatus is provided which moves the
focused laser beam in a circle or other path which controls dosimetry of radiation

1 applied to a target area, with the scanning rate being adjusted to limit thermal
diffusion from the irradiated target site for selective target specific energy
deposition. In one embodiment, a hand piece is used in which the scanning is
5 provided by rotating optics within the hand piece, with the speed of rotation
determining the scanning rate. In an alternative embodiment the scanning may be
provided by a rigidly mounted head containing the scanning optics. When used for
dermatologic purposes, the adjustable scanning mechanism prevents radiation from
10 impinging on tissue for more than about one millisecond in one embodiment for the
selective necrosis of highly-filled port wine stain blood vessels, while leaving
adjacent tissue undamaged. The dwell time of the laser beam is designed to match
the thermal diffusion time for destruction of the wall of the abnormal vessels, and
15 some surrounding collagen with the dwell time adjusted by the scanning rate.
Neither the small, empty normal vessels nor the collagen adjacent the normal
vessels are attacked by the impinging radiation since they contain no absorbing
chromophore. In one embodiment, a CW laser is used, with the time per scan
20 being maintained at less than 60 milliseconds to give the visual impression of a
continuous ring of light to the operator, with a rotary scan diameter of 2.8
millimeters and a focal spot size of 0.14 millimeters. The wavelength of the laser
25 is held below 600 micrometers so that the hemoglobin in the erythrocytes absorb
sufficient radiation for the necrosis of the vessels containing the erythrocytes.
While the subject invention will be described in connection with CW lasers, pulsed
lasers may be used and are within the scope of this invention.
30

In one embodiment the simple hand piece for a CW laser is provided with
cooling apparatus involving a cooling ejectant line and a cooling suction line on
opposite sides of the nose of the hand-held device. In a further embodiment,
35 bleachable indicators are painted onto the skin prior to usage to indicate to the
operator what areas of the epidermis have been irradiated.

1 It has also been found that to reduce the chances for regeneration of a
 lesion it is possible to slightly lengthen the pulse which is delivered to the given
 target area extending the thermal damage into the collagen immediately sur-
 5 rounding the abnormal vessel. An exposure time of about one millisecond allows
 heating as far as 20 micrometers from the edge of the abnormal vessel. This is
 adequate to insure total vessel destruction but is still small compared with the
 spacing between vessels, insuring sparing of a significant volume of the dermis.
 .0 The exact exposure time involves the correlation between the length of thermal
 diffusion damage and redevelopment or final destruction of the lesion. Other
 considerations include the fluence level of the focused light required to produce
 15 the desired damage effect and the depth of field, it being understood that the
 target tissue must be raised to an adequate temperature for a sufficient length of
 time to produce thermal necrosis.

In order to achieve adequate fluence levels from available laser systems, the
 20 laser beam diameter and depth of field are controlled by any of a variety of
 focusing techniques. The depth of field is made large enough to compensate for
 any variation in the distance between the target and the delivery system whether
 this distance is maintained by a mechanical fixture or simply by the operator's
 25 hand. Note that a larger depth of field makes the control of the distance between
 the delivery system and the skin less difficult. The larger depth of field is also
 useful because, when the skin is depressed by as little as one millimeter by the nose
 of the tool, this can cause a 1 millimeter bulge inside the nose of the instrument.
 30

Other uses for the subject invention include the treatment of other
 cutaneous vascular lesions such as telangiectasias, nevus araneus, strawberry
 nevus, cavernous hemangiomas, cherry hemangiomas, and venous lakes. The
 35 treatment of deep, cavernous hemangiomas is expected to require multiple
 treatments since these lesions are so hypertrophic that vessels nearest to the

1 surface will optically shield underlying vessels. By causing thermal necrosis of the
surface vessels and permitting adequate time for phagocytic removal of the
necrotic tissue, the next layer of vessels is made vulnerable to the laser
5 irradiation. It should be noted that laser wavelength may also be a more important
consideration in treating very deep lesions since very strongly absorbed wave-
lengths will provide shallower depth of necrosis per treatment. Another use for the
invention is in the treatment of psoriasis. Psoriasis is characterized by an
abnormal, ectatic and hypertrophic vasculature, along with other abnormal
features. Selective destruction of the vascular component permits destruction of
the lesion without subsequent tissue inflammation and psoriatic regeneration.

The device is also useful in the treatment of various forms of neovascular-
ization of the eye including diabetic retinopathy and senile macular degeneracy.
For this type of lesion a microscopic rather than hand held delivery system is
required. In the case of retinal disorders, where melanin absorption in the retinal
pigment epithelium could cause undesirable remote heating, a line scan, operator
defined and computer controlled, allowing the scan to exactly follow the vascular
line would be preferable. However, even full area scanning would destroy less
tissue than present modalities for treating each of these diseases. The currently
accepted modality calls for intentional destruction of large tissue volumes to
reduce the production of angiogenic substances. The scanner described herein,
however, has the potential to limit damage to such small volumes that frequently
repeated treatments, with greater preservation of visual acuity may be possible.

Another use for the device is in the treatment of structures bearing
melanosomes, including actinic keratoses, lentigo, malignant melanomas, and
freckles. In this case the dwell time (i.e., scanning rate) is adjusted to destroy the
melanosome bearing cells, leaving adjacent cells unharmed.

BRIEF DESCRIPTION OF THE DRAWINGS

These and other features of the subject invention will be better understood in connection with the detailed description taken in conjunction of which:

Figure 1 is diagrammatic representation of the utilization of the scanning hand piece in the treatment of a typical port wine stain;

Figure 2 is a diagrammatic illustration of the operation of the hand piece of Figure 1 illustrating a focal spot which moves around in a circular scan within the nose of the hand piece while the hand piece is moved across the area to be irradiated, liquid cooling and suction pipes being utilized to provide fluid which cools the irradiated area;

Figure 3A is a cross-sectional and diagrammatic illustration of one embodiment of the hand piece in which a rotating offset lens is utilized to scan the focused beam in a circular pattern;

Figure 3B is a diagrammatic and cross-sectional illustration of an alternative device utilized for the scanning of the beam by providing a rotating optical wedge interposed in the optical path;

Figure 4 is a top view and schematic diagram of the circular scan produced by the hand pieces of Figure 3A and Figure 3B;

Figure 5 is a diagrammatic representation of the result of irradiation of a port wine stain with the prior art ultrashort laser pulses, illustrating the rupturing of the port wine stain blood vessels causing the spewing forth of blood;

Figure 6 is a diagrammatic representation of the result of irradiation of a port wine stain with a prior art CW or long pulse laser showing complete necrosis of all irradiated areas, and;

Figure 7 is a diagrammatic representation of the result of utilizing the subject scanning system in which thermal diffusion is limited, causing necrosis of

1 the port wine stain blood vessel, its wall and a very small portion of the collagen
immediately adjacent the vessel wall, without disruption of the majority of the
normal tissue within the port wine stain area.

DETAILED DESCRIPTION

Referring now to Figure 1, in one embodiment a patient 10 having a port
wine stain 12 is being treated by laser radiation from a laser source 14 which is
channeled by fiber optic cable 16 to a hand-held unit 20 which has therein internal
optics utilized to provide scanning of a beam within the nose portion 22 of the tool.
This is accomplished in one embodiment through drive motor 24 utilized to control
the scan speed by rotating cable 26 which drives a hollow cylindrical barrel in the
tool that carries the focusing optics. Alternatively, air drive motors or small
electrical motors may be used in the hand piece to drive the rotating optics.
Hollow shaft motors (either electrically or pneumatically driven) may incorporate
the optical path within the hollow shaft, supporting the rotating element on the end
of the shaft. Solid shaft motors must be used with gear or other coupling
mechanisms to drive the rotating element. An optional source of cooling liquid 28
is applied to the hand tool which is channeled to the nose portion 22 and is removed
by a suction unit 30 such that the area of the target adjacent the nose of the tool is
cooled.

The operation of the hand tool can be better seen in conjunction with
Figure 2 in which like reference characters are utilized between Figures 1 and 2.
In Figure 2 an optical system 32 is utilized to focus and rotate a focal spot 34 such
that in the illustrated embodiment the spot rotates in a circle 36 as illustrated by
the dotted arrow. Cooling fluid is delivered at one side of nose 22 by a delivery
tube 38 and is removed by a suction tube 40 as illustrated.

1 It is the purpose of the rotating optics within the hand tool to scan the focal
spot such that it resides over a target for no longer than about one millisecond in
one embodiment. The control of the scan rate controls the time that the focused
5 spot resides at a given location within the target area and is readily adjustable by
the scan rate. While a circular scan is illustrated in the embodiment of Figure 2, it
will be appreciated that raster scan, Rosette type, orbital, elliptical, or other scan
patterns scans may be performed by optics to prevent the focal spot from residing
10 at any given location for longer than a predetermined period of time. In one
embodiment the depth of field is made greater than two millimeters by virtue of
the focal system aperture utilized. In this embodiment, a focal spot size of 0.14
15 millimeters, a scan pattern diameter of 2.8 millimeters, and a scan rate of less
than 60 milliseconds per cycle are used. For port wine stains, the wavelength of
the laser is held below 600 micrometers so that the hemoglobin in the erythrocytes
absorbs sufficient radiation to provide for the necrosis of the vessels containing the
20 chromophores.

In one operative mode, the hand piece is moved in a serpentine fashion as
illustrated by dotted arrow 33 across an area 46 which corresponds to the area of
25 the port wine stain.

As mentioned before, a dye may be first applied to the affected area which
changes color upon irradiation by focal spot 34 such that the treated area may be
ascertained with a high degree of certainty. This aids the operator who may be
30 unable "see" which areas have been treated since the treatment is so gentle as to
provide minimal visible color change of the lesion. The actual lightening of the
lesion occurs slowly, over a period of days to weeks as the body phagocytizes the
necrotic tissue. Instead of a hand held unit, the same treatment may be provided
35 by a programmed scanner. In such a case the indicating dye would not be
necessary.

Referring to Figure 3A, hand tool 20 may take on a configuration in which a body 50 includes a hollow cylindrical barrel illustrated at 52 to be rotated via cable 26 or other means about an axis 54 which is typically the central axis of the hand tool. The barrel is supported via a bearing system generally indicated by bearings 56.

Laser radiation is transmitted to the hand tool via fiber optic cable 16 which is coupled via a cable termination 60 to a lens 62 which collimates the light generally along axis 54. A lens 64 having a convex surface 66 focuses the parallel light as indicated by dotted lines 70 to a point 72 on the surface of skin 74. Since the optical axis of the lens is offset from centerline 54, its rotation via a barrel 52 causes the focal spot 72 to rotate on the surface of the skin 74. As can be seen, cooling liquid may be introduced through tube 38 such that liquid proceeds across the irradiated areas illustrated by 76 to suction tube 40.

The same system may include as a scanning means an optical wedge 80 which is shown in Figure 3B in which like apparatus is given like reference characters vis-a-vis Figure 3A. Note that power to rotate each of the barrels of Figures 3A and 3B is delivered by line 26, be it mechanical, electrical, hydraulic, or pneumatic. Note also that a fixed lens 82 is provided in nose 22 of the hand tool to achieve focusing.

Indeed, any focusing optics which is moved so as to provide a scanning beam, be it a raster scan, a circular scan, a line scan, or an elliptical scan, or some combination of these is within the scope of this invention. It is only important that the focal spot 72 not remain over any point within a target area 74 for any longer than is necessary for the particular purpose intended. For port wine stains this means that the dwell time for the spot should be on the order of one millisecond in order to prevent the type of damage which will now be described.

1 Prior to describing the damage done by thermal diffusion for radiation
impinging upon the skin for too long a period of time and referring now to Figure 5,
when using ultrashort pulses to treat port wine stains, it will be appreciated that
5 the abnormal, ectatic vessels are those illustrated by reference character 92. Each
of these vessels is nearly completely filled with blood and has a wall 94 thickness
of approximately 1-2 microns. Normal blood cells are indicated by reference
character 96 and are shown to be approximately 1/10 the size of the inflated port
10 wine vessels, with the spacing between the vessels being approximately 100
microns and with the diameter of the port wine stain vessels being 50 microns,
whereas the average diameter of the normal vessels is approximately 5 microns.
15 This is, of course a highly schematic illustration, using "average" values. Actual
vessel sizes and spacings vary widely.

What happens with the ultra-short pulses is that the thermal diffusion as
illustrated by arrows 102 is not sufficiently long to necrose both the blood in the
20 port wine stain vessel as well as the vessel wall. Rather what happens is that the
vessel wall is ruptured due to the ultrashort pulse thereby bursting the vessel and
causing blood to spurt out as illustrated at 104. This is highly undesirable due
primarily to the fact that the blood spurting through the surrounding tissue here
25 illustrated by reference character 100, interferes with further irradiation. In
addition this point by point pulsed irradiation is very tedious for the operator.

Referring to Figure 6, CW or a long pulse radiation causes a necrosis not
30 only of the blood within the port wine stain vessels 92 but also necroses the
collagen between the vessels as well as normal vessels 96. For example, a 50 msec
pulse gives 170 micron thermal diffusion length from each absorbing vessel, such
that there is an overlap in the manner illustrated, with the result that the entire
35 irradiated area is damaged as shown by the shading.

1 Referring now to Figure 7, when the dwell time for the radiation is limited
to approximately one millisecond in the subject system, the area necrosed is
indicated to include the port wine stain vessel 92 and extends approximately 20
5 micrometers (microns) into the tissue immediately adjacent the port wine stain
vessels. It will be noted that little of surrounding tissue 100 and few of the normal
vessels 96 are affected. The subject technique thus leads to a relatively painless,
treatment for the port wine stain syndrome, without extensive tissue damage,
0 sloughing, and scab formation.

Damage induced by thermal diffusion as illustrated by wavy arrows 102 may
be controlled in length by viture of the spot size and the scanning time, which is
5 controllable to the extent necessary to preclude damage to the majority of the
avascular tissue and the normal blood vessels within the port wine stain. As
mentioned before, for a focal spot size of 0.14 millimeters and a scan diameter of
2.8 millimeters with a cycle time 60 milliseconds, even without the utilization of
0 external cooling, the necrosed area is limited to 20 microns from the walls of the
vessels which are enlarged and are filled with blood, whereas the vessels that are
normal and generally less than 10 percent of which are filled with blood at any
point in time, are virtually unaffected. Note that the aforementioned cooling may
5 be added for patient comfort, to avoid the use of anesthetic, and to increase the
specificity of the treatment.

It will be appreciated that one difference between the subject treatment
and that of Figure 5 is that the abnormal vessel is necrosed without the spurting
0 forth of blood which decreases the effectiveness of treatment. The present
technique is more effective than completely necrosing the whole port wine stain
area because the sloughing associated with the prior art technique of Fig. 6 is
5 eliminated. Thus a painful and time-consuming procedure is replaced through the
utilization of the control of the dwell time of the radiation impinging on the target

1 area to an extent not heretofore possible, thereby effectuating a treatment that
may be used without anesthetic and which achieves the desired result without large
amounts of scabbing and sloughing.

5 It will be appreciated that the control of laser radiation through the
scanning device described in this invention is applicable to other situations which
require a controlled amount of energy be delivered to a given absorbing target site.
0 Thus for instance the subject technique is applicable to treatment of any abnormal,
ectatic vasculature, any highly filled vasculature within tissue, any preferentially
stained (for laser absorption) target tissue whether or not embedded in normal
tissue to be spared. With appropriate choice of laser wavelength it may be used to
5 destroy non-vascular pigmented targets, in particular melanin bearing tissues such
as actinic keratosis, lentigo, malignant melanoma, or portions of the retinal
pigment epithelium.

Having above indicated a preferred embodiment of the present invention, it
) will occur to those skilled in the art that modifications and alternatives can be
practiced within the spirit of the invention. It is accordingly intended to define the
scope of the invention only as indicated in the following claims:

CLAIMS

What is claimed is:

1. Apparatus for use with a laser comprising a scanning mechanism which controls the dosimetry of radiation applied to a target area, said scanning mechanism being adjusted in scanning rate to limit thermal diffusion from the light absorbing portion of the radiated target site for selective target specific energy deposition.

2. The apparatus of claim 1 wherein said apparatus includes a hand piece having an apertured nose and wherein said apparatus further includes:

means including said laser for generating laser radiation;

means including fiber optic means for coupling radiation from said laser to said hand piece; and

means within said hand piece for focusing the radiation from the output of said fiber optic means and for providing a scanning pattern of said focused radiation at the nose of said hand piece.

3. The apparatus of claim 2 and further including means carried by said hand piece for cooling the area adjacent the nose of said hand piece.

4. The apparatus of claim 2 wherein said scanning means includes means for inducing a rotary scan of the focused radiation from the output of said fiber optic means.

1 5. The apparatus of claim 4 wherein said rotary scan means includes a hollow
rotatable shaft having a centerline, said shaft being located within said hand piece,
and means for rotating said shaft, said focusing means being mounted to said
5 hollow cylindrical shaft such that said focusing means is offset from the centerline
of said hollow cylindrical shaft so as to provide for a rotary scan of the focused
radiation from said laser, the dwell time of the focal spot at the nose of said hand
piece being controlled by the speed of rotation of said hollow rotational shaft.

10

6. The apparatus of claim 5 wherein said offset focusing means includes a lens
offset from the centerline of said rotatable shaft.

15

7. The apparatus of claim 4 wherein said rotary scan means includes a hollow
rotatable shaft within said hand piece, and further including an optical wedge
mounted to said hollow rotatable shaft, said focusing means including a fixed lens
20 in the nose of said hand piece.

20

8. The apparatus of claim 1 wherein said scanning mechanism includes means
for focusing radiation from said laser to produce a focal spot and wherein said
25 scanning rate is adjusted such that the dwell time of said focal spot is no longer
than the time required to damage the tissue targeted for destruction.

25

9. The apparatus of claim 1 and further including means for adjusting the dwell
30 time of the radiation applied to said target area to be less than the time required
to damage the tissue targeted for destruction.

30

35 10. A method for the treatment of port wine stains comprising irradiating the
port wine stain with focused laser radiation which is scanned in a pattern which

35

1 limits the dwell time of the focused radiation on the area to a time sufficient to
selectively necrose highly filled blood vessels, while leaving adjacent tissue and
empty blood vessels undamaged.

5

11. The method of claim 10 wherein said scanning includes providing a rotary
scanned beam of laser energy which is focused to a focal spot and which is caused
to scan in a predetermined pattern across the port wine stain such that the focal
spot does not dwell on any target area for any more than 1 millisecond.

10

12. The method of claim 10 wherein said scanning is accomplished at a rate
which limits the thermal diffusion length of heat generated when said focused laser
radiation impinges on the port wine stain.

15

13. The method of claim 12 wherein said thermal diffusion length is limited to a
distance corresponding to the average wall thickness of a highly filled blood vessel
and a predetermined small amount of surrounding tissue.

20

14. The method of claim 13 wherein said predetermined amount of collagen
exists no more than 20 microns away from the outside wall of a highly filled blood
vessel.

25

15. The method of claim 10 wherein said radiation is generated by a CW laser.

30

16. The method of claim 10 wherein the dwell time of the focused laser
radiation is designed to match the thermal diffusion time for destruction of the
wall of abnormal vessels and a small portion of the surrounding collagen, with the
dwell time being adjusted by the scanning rate.

35

1 17. The method of claim 10 wherein a CW laser is used, wherein a rotary scan is
used with a time per scan being maintained less than 60 milliseconds, with the
rotary scan diameter of the focused radiation being 2.8 millimeters and wherein,
5 the focused radiation forms a spot having size of 0.14 millimeters, the wavelength
of the laser being below 600 micrometers.

10 18. A method for the selective destruction of a vascular component comprising
radiating a target area on the human body with a focused beam of laser radiation,
which beam of laser radiation is scanned across the area in such a manner that the
dwell time of the focused beam is controlled such that the focused radiation
15 produces a thermal diffusion which is limited by the scan rate, thereby providing
selective necrosis of predetermined components of the human body while leaving
other components of the human body undisturbed.

20

25

30

35

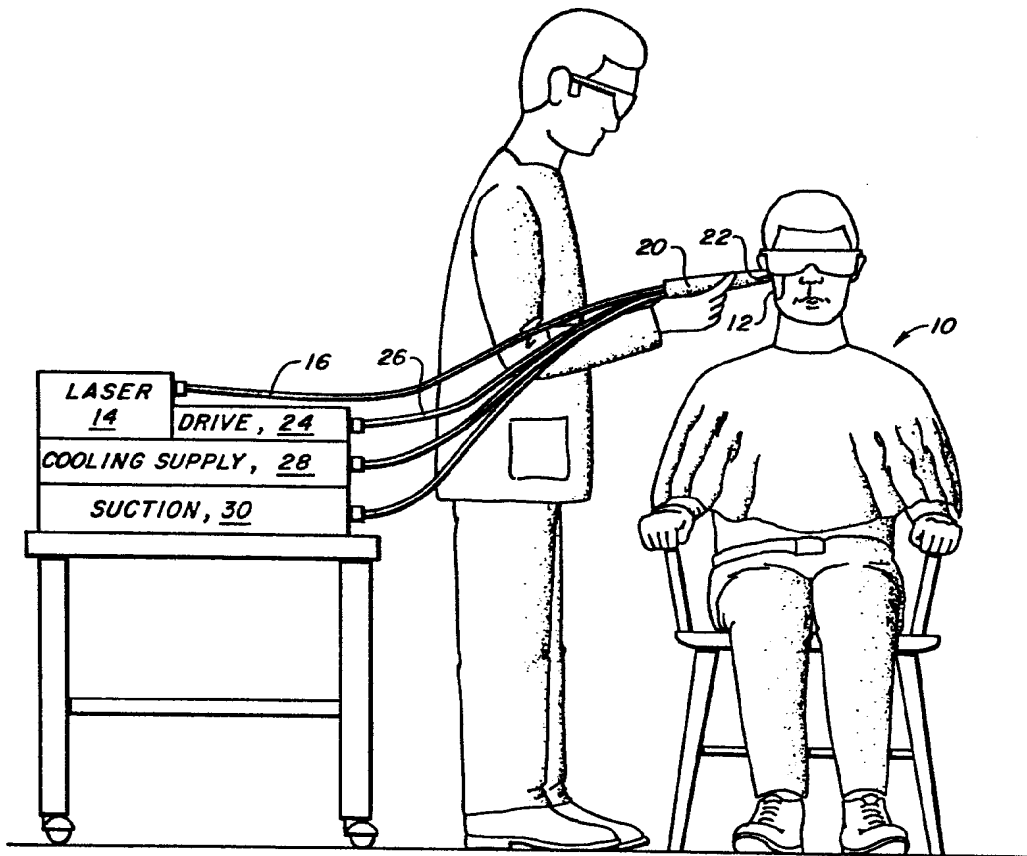


FIG. 1

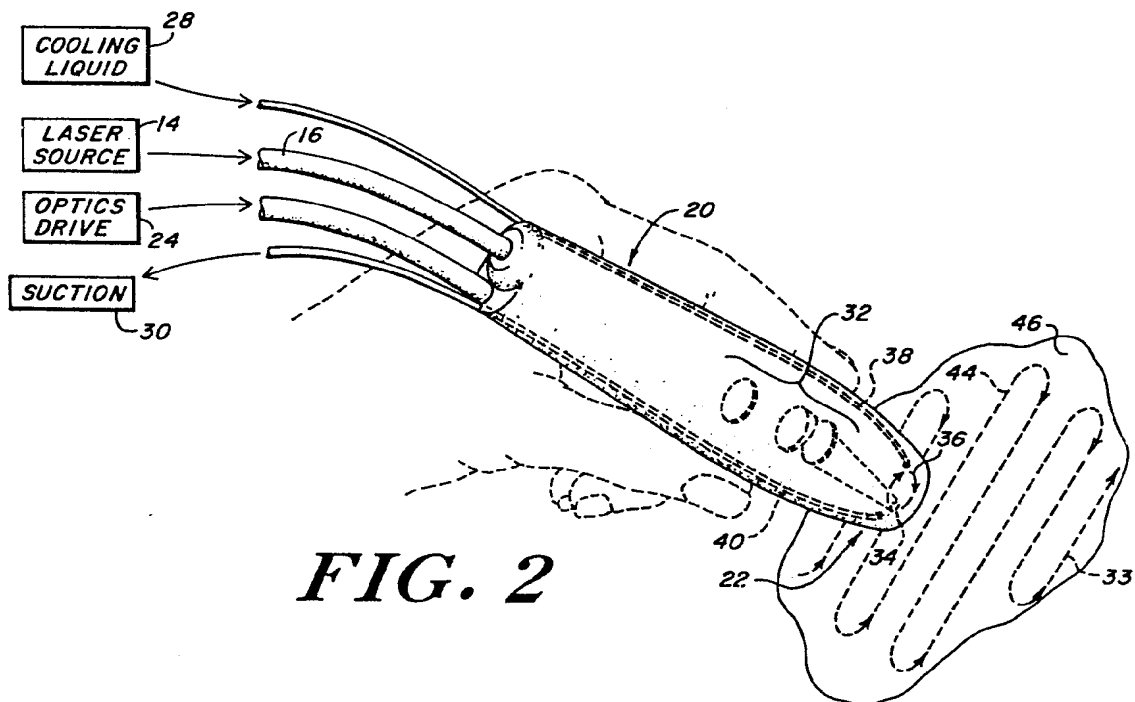
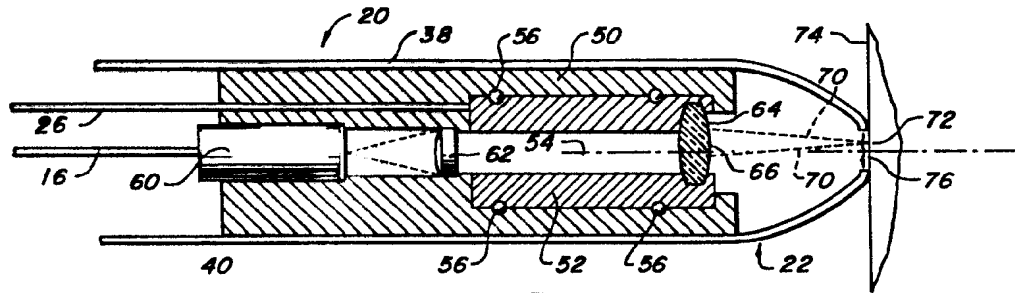
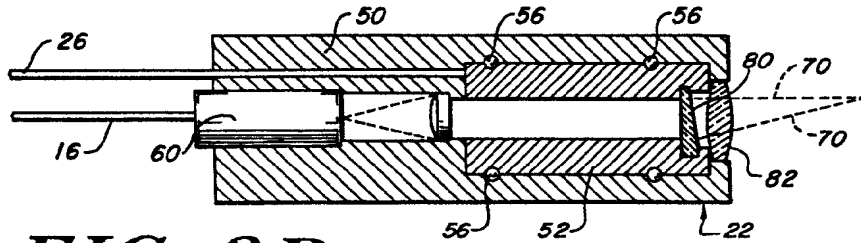
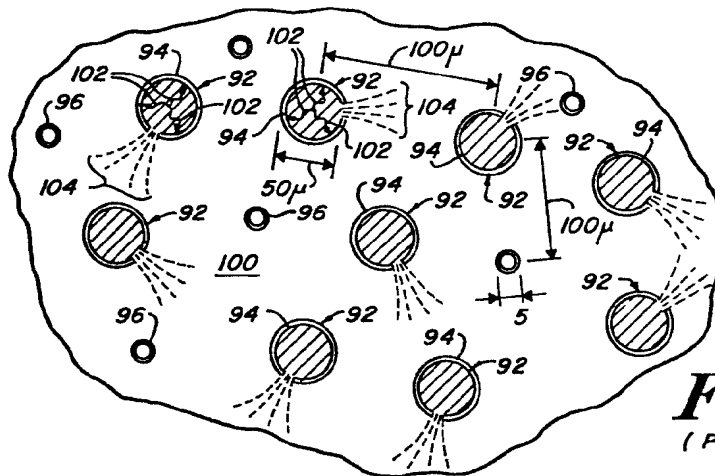
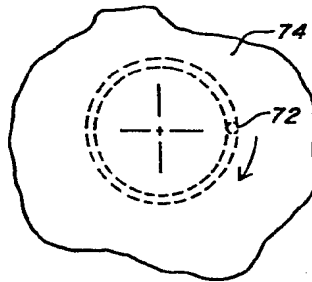


FIG. 2

**FIG. 3A****FIG. 3B****FIG. 4****FIG. 5**
(PRIOR ART)

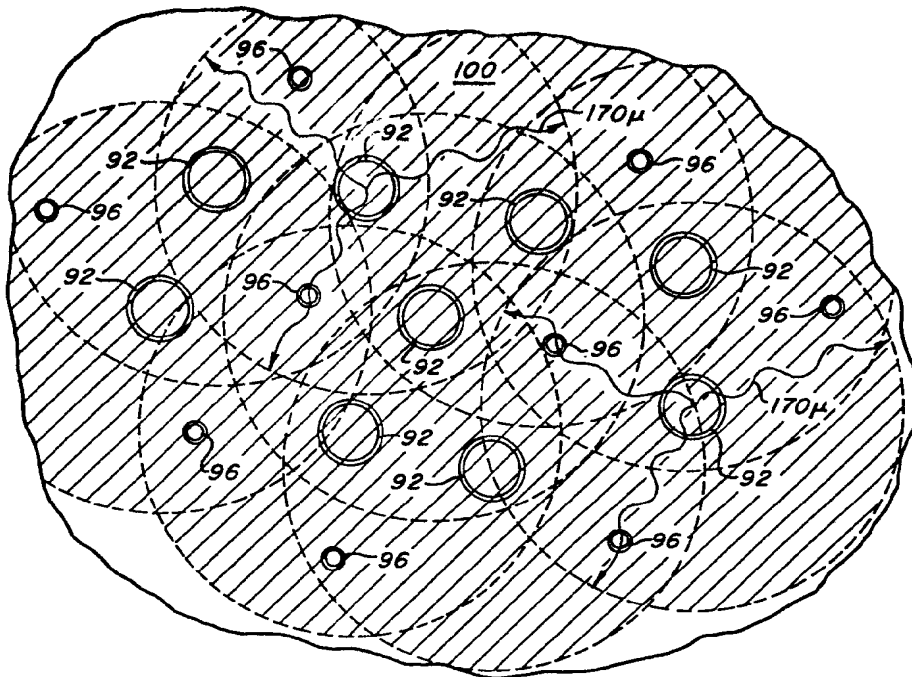


FIG. 6
(PRIOR ART)

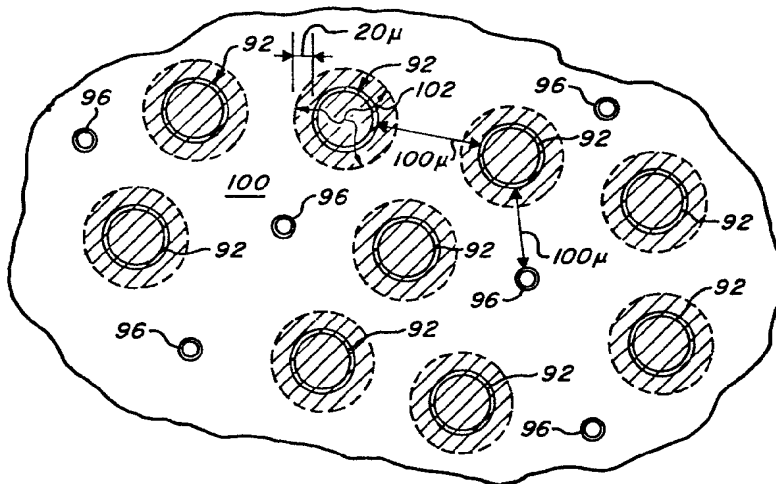


FIG. 7



European Patent
Office

PARTIAL EUROPEAN SEARCH REPORT
which under Rule 45 of the European Patent Convention
shall be considered, for the purposes of subsequent
proceedings, as the European search report

0172490

Application number

DOCUMENTS CONSIDERED TO BE RELEVANT			EP 85109894.7
Category	Citation of document with indication, where appropriate, of relevant passages	Relevant to claim	CLASSIFICATION OF THE APPLICATION (Int. Cl. 4)
X,D	US - A - 4 069 823 (ISAKOV) * Abstract; column 2, line 61 - column 3, line 30; column 3, line 59 - column 4, line 19; fig. 3,5 *	1,2	A 61 B 17/36 A 61 N 5/06
	--		
X,D	US - A - 4 316 467 (MUCKERHEIDE) * Abstract; column 1, line 65 - column 2, line 19; fig. 1 *	1,2	
	--		
X	EP - A2 - 0 075 912 (HITACHI) * Abstract; page 3, line 13 - page 4, line 5; fig. 2,6,13, 15 *	1,2	
	--		
P,A	EP - A2 - 0 130 950 (TESSIORE) * Abstract; page 2, last paragraph - page 3, first paragraph; page 4, line 22 - page 5, line 19 *	1,2,4	TECHNICAL FIELDS SEARCHED (Int. Cl. 4) A 61 B A 61 N
INCOMPLETE SEARCH --			
<p>The Search Division considers that the present European patent application does not comply with the provisions of the European Patent Convention to such an extent that it is not possible to carry out a meaningful search into the state of the art on the basis of some of the claims</p> <p>Claims searched completely: 1-9 Claims searched incompletely: -- Claims not searched: 10-18 Reason for the limitation of the search:</p> <p>according to Art. 52(4)</p>			
Place of search VIENNA		Date of completion of the search 27-11-1985	Examiner NEGWER
<p>CATEGORY OF CITED DOCUMENTS</p> <p>X : particularly relevant if taken alone Y : particularly relevant if combined with another document of the same category A : technological background O : non-written disclosure P : intermediate document</p> <p>T : theory or principle underlying the invention E : earlier patent document, but published on, or after the filing date D : document cited in the application L : document cited for other reasons & : member of the same patent family, corresponding document</p>			



DOCUMENTS CONSIDERED TO BE RELEVANT			CLASSIFICATION OF THE APPLICATION (Int. Cl. 4)
Category	Citation of document with indication, where appropriate, of relevant passages	Relevant to claim	
A	<u>EP - A1 - 0 074 914 (SERIEL)</u> * Page 9; claims 1,2 * --	1,2,4	
A	<u>US - A - 3 653 384 (SWOPE)</u> * Abstract; column 1, line 66 - column 2, line 8 * --	1,2,4, 5	
A	<u>US - A - 3 821 510 (MUNCHERYAN)</u> * Abstract; column 2, lines 48-66 * --	1-3	
X	<u>DE - A1 - 2 829 516 (MESSERSCHMITT)</u> * Page 5, line 8 - page 6, line 13; fig. 4,5,7 * --	1,2	TECHNICAL FIELDS SEARCHED (Int. Cl. 4)
X	<u>DE - A1 - 3 245 846 (MITSUBISHI)</u> * Abstract; page 1, claim 1; fig. 1,3,6 * ----	1,2	

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Device and system for biostimulation of tissue.

(57)

A device for biostimulation of tissue is disclosed comprising an array of substantially monochromatic radiation sources of a plurality of wavelengths, preferably of at least three different wavelengths. The radiation sources are arranged within the array such that radiation of at least two different wavelengths passes directly or indirectly through a single point located within the treated tissue. The radiation sources are preferably laser diodes, superluminous diodes or similar light-emitting diodes that, while low-power radiation sources, can provide significant energy densities to a treatment area. A method of treatment of tissue comprising exposing the treated tissue to the above-described device is also disclosed. The device for biostimulation of tissue may be included within a system with a control panel (20), a power source (40), means (47) for varying pulse frequency, means for varying pulse duration, means (72) for timing the period of treatment, means (75) for measuring the conductivity of the treated tissue, means for measuring the optical power emitted by the radiation sources and/or means (30) for detecting emissions from the radiation sources.

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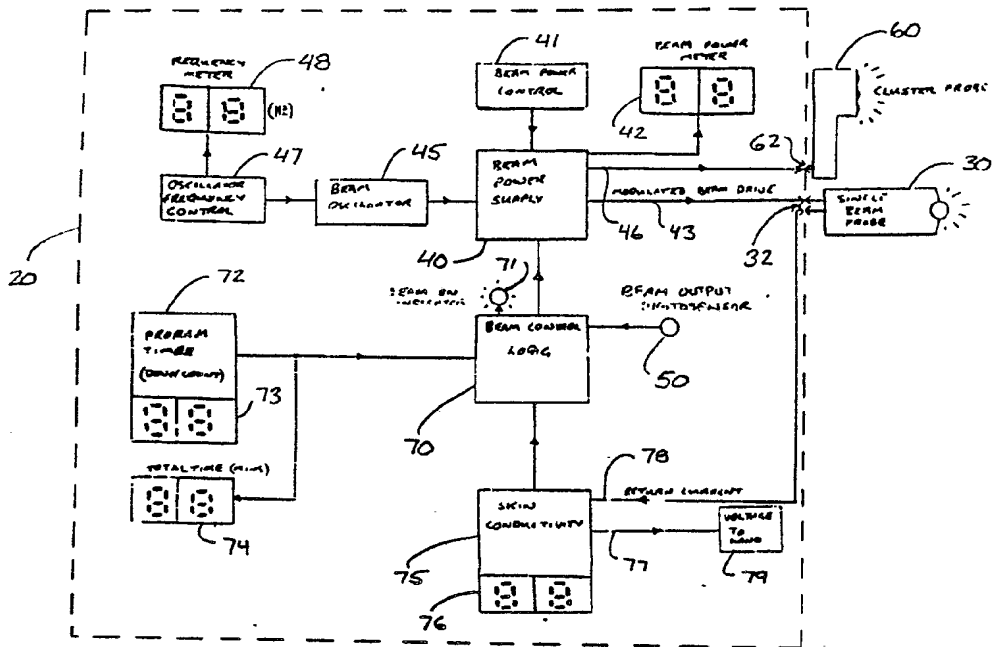


FIG. 4

A DEVICE FOR BIOSTIMULATION OF TISSUE AND METHOD FOR TREATMENT OF TISSUE

FIELD OF THE INVENTION

5 This invention relates to a new and improved device, method and system for biostimulation of tissue with low-power radiation, preferably substantially monochromatic radiation, having a plurality of wavelengths and having significant power densities over a treatment area.

BACKGROUND OF THE INVENTION

10 For many years, high-powered, highly focused lasers have been widely used to cut and destroy tissue in many surgical techniques. More recently, low-powered lasers, less sharply focused, which do not sever or destroy tissue have been found or are thought to effect numerous metabolic processes, including cell
15 division, cyclic-AMP metabolism, oxidative phosphorylation, hemoglobin, collagen and other protein synthesis, leukocyte activity, tumor growth, production of macrophage cells and wound healing. See, for example, Karu and Letokhov "Biological Action of Low-Intensity Monochromatic Light in the Visible Range" in Laser Photobiology and Photomedicine, ed. Martellucci, p. 57-66 (Plenum Press 1985); Passarella, et al., "Certain
20 Aspects of Helium-Neon Laser Irradiation on Biological Systems in Vitro" in Laser Photobiology and Photomedicine, ed. Martellucci p. 67-74 (Plenum Press 1985); see generally, Parrish, "Photomedicine: Potentials for Lasers. An Overview," in Lasers in Photomedicine and Photobiology, ed. Pratesi, p. 2-22 (Springer 1980); Giese, "Basic Photobiology and Open Problems" in Lasers in Photomedicine and Photobiology, ed. Pratesi, p. 26-39 (Springer 1980); Jori, "The Molecular Biology of Photodynamic Action"
25 in Lasers in Photomedicine and Photobiology, ed. Pratesi, p. 58-66 (Springer 1980). Although the precise mechanism for these effects is not fully understood, it is believed to be tied to the activity of specific wavelengths of radiation in or near the range of visible light. Infrared laser radiation has been shown to increase ATP concentration and ATPase activity in living tissues. Bognani, et al., "Effects of GaAs Pulsed Lasers on ATP Concentration and ATPase Activity In Vitro and In Vivo", International Cong. on Lasers in
30 Medicine and Surgery, p. 47 (1985).

Radiation sources operating in or near the range of visible light, including lasers, emit photons which may interact with biological molecules to produce photochemical reactions and subsequent biologic effects. Photochemical and photobiological events at the atomic level depend upon the wavelength of radiation used to cause such events and occur without regard to the source of photons. However, the molecular effects,
35 kinetics and products can be quantitatively and qualitatively altered one or more by other properties of radiation sources, e.g., monochromaticity, coherence and high power and energy density.

Most forms of photoexcitation are "quantum specific," i.e., excitation will only occur if a bundle of energy of a precise quantity is present to excite a given molecule or part of a molecule. A photon has energy E according to the formula:

$$E = h \times f = \frac{h \times c}{\text{wavelength}}$$

45 where f is frequency, h is Planck's constant and c is the speed of light. If a photon having a quantum of too little or too much energy is directed at a target molecule, it may not be absorbed; the photon must be of an exact energy to have an effect.

Only radiation which is absorbed has photochemical effects. X-rays, gamma rays and other absorbed high-energy photons affect human tissues by relatively indiscriminate ionization of molecules. The ionized
50 molecules are highly reactive and covalent bonds may be broken or formed. Infrared photons excite specific vibrational or rotational modes in specific target molecules. The quantum of energy required to produce vibrational or rotational excitation is dependant on the character (e.g., double bond vs. ring structure) and location (e.g., near an electrophilic group vs. near a nucleophilic group) of the molecule. While it is believed that infrared photons may affect specific biological processes or transformations, the

most significant biological effect of these wavelengths is probably the heating caused by dissipation of the vibrational and rotational energy, which can significantly effect biological reactions in the vicinity of the dissipating molecule. The energy of photons in the ultraviolet and visible wavelengths causes electronic excitation of specific chromophores (i.e., molecules that absorb a photon of a given wavelength and use the energy to cause transition of an electron to a higher energy state). The decay of these stimulated molecules can then lead to specific reactions, including emission of a new photon, transfer of an electron or dissipation of heat.

In the past it has been difficult, however, to expose more than the first few layers of human skin or tissue to visible (400-700 nm) and ultraviolet (200-400 nm) radiation. Pigments and other molecules in the outer layers of skin are known to absorb the majority of visible and ultraviolet radiation, as shown in Figs. 1-3. Table 1 summarizes the approximate penetration of various wavelengths of radiation into the skin.

Table 1

Approximate Depth of Penetration of Optical Radiation in Fair Caucasian Skin to a Value of $1/e$ (37%) of the Incident Energy Density	
Wavelength, nm	Depth, nm
250	2
280	1.5
300	6
350	60
400	90
450	150
500	230
600	550
700	750
800	1200
1000	1600
1200	2200

As shown in Fig. 3, no ultraviolet radiation and approximately only 5% of most visible radiation penetrates to the subcutaneous layer of the skin. As a result, applying visible and ultraviolet radiation to the skin has little or no effect upon target molecules in lower layers that would become stimulated if exposed to those wavelengths of radiation.

While higher powered radiation sources can deliver greater energy to deeper layers, it is undesirable to directly expose tissue to large amounts of ultraviolet radiation due to the adverse effects of such radiation upon some molecules and cellular functions, e.g., DNA can be "mutated" by ultraviolet radiation.

It would therefore be desirable to provide a safe device and method for biostimulation of tissue that will stimulate biological processes affected by visible red and infrared radiation and also stimulate biological processes in lower layers of tissue that are affected by ultraviolet and visible radiation and would normally be inaccessible to radiation applied to the surface of the tissue because of the absorption of visible and ultraviolet radiation by skin pigments and other molecules.

SUMMARY OF THE INVENTION

A device for biostimulation of tissue is disclosed comprising an array of substantially monochromatic radiation sources of high power density and a plurality of wavelengths, preferably of at least three different wavelengths. The radiation sources are arranged within the array such that radiation of at least two different

wavelengths passes directly or indirectly through a single point located within the treatment target tissue. The radiation sources are preferably laser diodes, superluminous diodes or similar light-emitting diodes. A method of treatment of tissue comprising exposing the treated tissue to such a device is also disclosed. The device for biostimulation of tissue may be included within a system with a control panel, a power source, means for varying radiation pulse frequency, means for varying radiation pulse duration, means for timing the period of treatment, means for measuring the conductivity of the treatment target tissue, means for measuring the optical power emitted by the radiation sources and/or means for detecting emissions from the radiation sources.

It is an object of the present invention to provide a device and method for low power radiation for biostimulation of tissue that can deliver higher energy densities and a greater number of potentially biostimulative photons to deeper tissue layers.

It is another object of the invention to provide a device and method for biostimulation of tissue with multiple wavelengths of radiation.

It is a further object of the invention to provide a device and method for biostimulation of tissue utilizing semiconductor laser, superluminous or light emitting diodes as sources for low-power radiation in the infrared, visible and/or ultraviolet frequency ranges.

BRIEF DESCRIPTION OF THE FIGURES

Fig. 1 is a graph summarizing the ultraviolet absorption spectra of major epidermal chromophores: DOPA-melanin, 1.5 mg% in H₂O; urocanic acid 10^{-4} M in H₂O; calf thymus DNA, 10 mg% in H₂O (pH 4.5); tryptophan 2×10^{-4} M (pH 7); tyrosine, 2×10^{-4} (pH 7). [From Pratesi and Sacchi, Eds., Lasers in Photomedicine and Photobiology, p. 165 (Springer 1980)].

Fig 2 is a graph summarizing the visible light absorption spectra of major human skin pigments. Parentheses indicate solvents used. [From Pratesi and Sacchi, Eds., Lasers in Photomedicine and Photobiology, p. 172 (Springer 1980)].

Fig. 3 is a pictorial and graphic representation of the relative penetration of various radiation wavelengths into human skin. [From D. Slimey and M. Wolbarsht, Safety With Lasers and Other Optical Sources (1980)].

Fig. 4 is a function block diagram of the device and system of the present invention.

Fig. 5 is a side view of a cluster probe or radiation source array used in the present invention.

Fig. 6 is a plan view of a radiation source array used in the present invention.

Fig. 7 is a plan view of another radiation source array comprising an alternative embodiment of the present invention.

Fig. 8 is a plan view of another radiation source array comprising an alternative embodiment of the present invention.

Fig. 9 is a simplified schematic view of the radiation beams of three diodes as used in the present invention impinging on a treatment tissue target.

DETAILED DESCRIPTION OF THE INVENTION

A. System Overview.

The block diagram of Fig. 4 shows the overall structure of the device and system of the present invention. A control unit 20 contains both controls and displays to read out control settings and measured values. To this control unit 20 a single beam probe 30 and a cluster probe 60 may be connected at plug-in connections 32 and 62, respectively. Central to the control unit 20 are a beam power supply 40 and, connected thereto a beam control logic unit 70. The beam power supply has two output lines 43 and 46. Output line 46 leads to plug-in connection 62 for the cluster probe 60. Output line 43 leads to plug-in connection 32 for single beam probe 30.

Beam power control 41 is connected to beam power supply 40 to permit setting of the beam power level. Beam power meter 42 is connected to the beam power supply 40 to show the power level being delivered.

To provide radiation beams modulated in pulses of various frequencies, a beam oscillator 45 is connected to the beam power supply 40. The beam oscillator 45 is controlled by an oscillator frequency control 47 connected to the beam oscillator 45. A frequency meter 48 connected to the oscillator frequency control 47 provides display of the selected beam modulation frequency. If pulse duration modulation is desired as a further form of modulation, this can be accomplished by additional refinements of the beam oscillator 45 and its corresponding oscillator frequency control 47.

The beam control logic unit 70 is, as previously noted, connected to the beam power supply 40. An indicator light 71 associated with the unit shows when the beam power is on. This is useful because the beam wavelengths being used may not be visible and treatment time, as well as power, is an important variable ($E = P \times T$). As a further check that the unit is functioning, the invention provides a beam output photosensor 50 connected to the beam control logic unit 70 described below). This photosensor 50 is sensitive to the frequencies of radiation produced by the unit and provides a signal when it receives radiation. To aid control of treatment time, a down-counting program timer 72 with a display 73 that follows the down count is connected to the beam control logic unit 70. A further display 75, also connected to the beam control logic unit 70, provides a display of total beam on-time (in minutes).

For some uses of the invention, it is desirable to locate body points of high skin conductivity. (These usually correspond to pain trigger points of inflammatory areas, as increased temperatures will raise skin conductivity.) The present invention does this with a skin conductivity measuring module 75 connected to beam control logic unit 70, with associated display 76. This module 75 delivers a small current (microamps) via a lead 77 connected to an electrode 79 held in the hand of a patient. The single beam probe 30 is used to form a return path from a selected skin location, utilizing lead 78 as a return current path (from plug-in connection 32) to the skin conductivity measuring module 75. If desired, the skin conductivity measurement can be used as a trigger for the beam control logic unit 70; that is, the beam control logic unit 70 can be set to enable a beam only when a preselected skin conductivity level is present. When this level is chosen to be very low, the beam is enabled whenever the return path probe is in contact with a skin area to be treated.

The present invention utilizes, as noted above, radiation in or near the infrared spectrum (above 700 nm), the visible light spectrum (400-700 nm) and the ultraviolet spectrum (200-400 nm). For convenience in the following, the radiation comprising the beams produced by the present invention may be referred to as "light", although it may be in the visible or ultraviolet spectra or in other nearby spectra.

The single beam probe 30 of the present invention is shaped like a fat pencil (Fig. 4). It emits radiation of a single frequency and is therefore of limited interest in connection with the present invention. The gravamen of the invention is use of multiple radiation sources of multiple frequencies. This radiation is emitted by the multiple radiation sources contained in cluster probes 60 used with the present invention.

Fig. 5 shows a side view of a cluster probe 60, having a thin cylindrical handle 61 and a thicker cylindrical head 62. Figs. 6 through 8 show three patterns of radiation sources that can be contained within the cylindrical head 62. The radiation is emitted from a plane very near one end-face of the cylinder 62. Figs. 6-8 show head-on views of several end-faces. As will be described below, the end-faces involve various configurations of radiation sources. Each of these configurations provides a different mix of radiation source frequencies and a somewhat different geometric configuration. These configurations accordingly produce different "mixtures" of radiation frequencies in the target tissue and different power densities.

The sources of light or radiation in each of the cluster probes 60 in Figs. 6-8, showing particular forms of radiation arrays, are semiconductor light emitting devices, e.g., light emitting diodes (LED'S). Two particular types of LED'S have been found most useful for purposes of the present invention: laser diodes and superluminous diodes. Laser diodes produce a beam of light or radiation that is essentially monochromatic, is sharply collimated and is coherent. That is, they produce light almost exclusively at one frequency (unless they are multi-mode type lasers) and the light beam has a small angle of divergence. Superluminous diodes are also used. These are similar but lack the coherence and the sharply monochromatic characteristics of laser diodes; yet they produce highly directional light that is also limited in its frequency range.

A number of commercially available semiconductor laser diodes exist. Typical of these are those described in "Optoelectronic Devices Data Book" published by Hitachi, Ltd. (September, 1984).

It has been found, however, that semiconductor laser diodes having somewhat higher power outputs and narrower beam divergence and spectral widths than the most widely manufactured components are also available and may enhance the advantages of the present invention. Not all frequencies are available in the range from ultraviolet through visible to infrared radiation. But enough are available that some selection among frequencies can be made. Among low power lasers suitable for the present invention, the laser

power rating (continuous power) of individual diodes is generally in the range from 5-500 milliwatts (mW). Laser diodes are available with continuous wave emission capability and as devices that must be pulsed. The following laser diode specifications have been found useful for the present invention:

1. Double Heterostructure Continuous Wave Laser Diode GaAlAs
 - 5 Wavelength: 750, 780, 800, 810, 820, 830, 850nm
 - Peak Power Output: 5mW - 500 mW (Class 3B)
 - Beam Divergence: 60 parallel, 12° perpendicular (typical, variable according to manufacturing method)
 - Polarisation: Linear 90-100%
 - Spectral Width: 0.02mm-1.0mm
- 10 2. Single Heterostructure Pulsed Laser Diode GaAs
 - Wavelength: 904nm
 - Peak Power Output: 70W
 - Avg. Power: 0.15-15mW (frequency dependent)
 - Max. Pulse Duration: 200 microsecs.
 - 15 Beam Divergence: 6°-15° parallel, 15°-30° perpendicular
 - Spectral Width; less than 3.5nm
3. Double Heterostructure Pulsed Laser Diode GaAs/GaAlAs Wavelength: 850-904nm
 - Peak Power: 325mW
 - Average Power: 40-80mW
 - 20 Faxed Duty Factor ($T_w \times F_n$): 15%; $f = 300\text{KHz}$, $T_w = 500$ microsecs.
 - Beam Divergence: 6°-15° parallel, 15°-30° perpendicular
 - Spectral Width: 2-3nm

As best seen in Fig. 6, the preferred embodiment of the cluster probe 60 of the present invention
 25 comprises an array 80 of five 660 nm superluminous diodes 166, one 820 nm laser diode 182, ten 880
 superluminous or laser diodes 188 and five 950 superluminous diodes 195. The diodes are arranged in a
 planar array such that the 820 nm diode 182 is positioned in the center of the array, five of the ten 880 nm
 diodes 188 are evenly positioned about the circumference of a circle of about 9.5 mm radius from the
 center of the array, the five 660 nm diodes 166 are evenly positioned about the circumference of a circle of
 30 about 18 mm radius from the center of the array such that a radial line from the center of the array to each
 660 nm diode 166 bisects the arc between two of the innermost 880 nm diodes 188, the 950 nm diodes
 195 are evenly positioned about the circumference of a circle of about 27 mm radius from the center of the
 array such that a radial line from the center of the array to each 950 nm diode 195 passes through the
 center of one of the innermost 880 nm diodes 188, and the remaining five 880 nm diodes 188 are evenly
 35 positioned about the circumference of a circle of about 27 mm radius from the center of the array such that
 a radial line from the center of the array to each outer 880 nm diode 188 passes through the center of one
 of the 660 nm diodes 166.

In an alternative embodiment shown in Fig. 7, the 820 nm diode, 182 is positioned in the center of the
 array 90, ten 880 nm diodes 188 are evenly positioned about the circumference of a circle of about 10.5
 40 mm radius from the center of the array, and five 660 nm diodes 166 and five 950 nm diodes 195 are
 alternately positioned and evenly spaced about the circumference of a circle of about 17 mm radius from the
 center of the array, such that a radial line from the center of the the array to each 660 nm or 950 nm diode,
 166, 195 respectively, bisects the arc between two of the 880 nm diodes 188.

Another alternative embodiment of the cluster probe 60 of the present invention, shown in Fig. 8,
 45 comprises an array 100 of ten 660 nm superluminous diodes 166, one 820 nm laser diode 182, ten 880 nm
 superluminous or laser diodes 188, and ten 950 nm superluminous diodes 195. The diodes are arranged in
 a planar array such that the single 820 nm diode 182 is positioned in the center of the array, ten 880 nm
 diodes 188 are evenly positioned about the circumference of a circle of about 9.5 mm radius from the
 center of the array, ten 660 nm diodes 166 are evenly positioned about the circumference of a circle of
 50 about 18 mm radius from the center of the array such that a radial line from the center of the array to each
 660 nm diode 166 bisects the arc between two of the 880 nm diodes 188, and ten 950 nm diodes 195 are
 evenly positioned about the circumference of a circle of 27 mm radius from the center of the array, such
 that a radial line from the center of the array to each 950 nm diode 195 passes through the center of one of
 the 880 nm diodes 188.

55

B. Theory of Operation

The diodes within the array of each embodiment are closely arranged such that although the radiation they produce is emitted in a narrow beam, their beams overlap a short distance away from the surface of the cluster probe 60. Thus, as can be seen in Fig. 9, two or more wavelengths of radiation from the array 60 simultaneously pass through some point in the tissue 100 being stimulated. In Fig. 9, the divergence of each radiation beam is schematically shown for three radiation sources 182, 188 and 195 assumed to line
 5 along a single line. Radiation source 182 is assumed to have a beam divergence of six degrees, while radiation sources 188 and 195 are assumed to have beam divergence of 15 degrees. No other optical effects such as reflection, refraction or scattering are assumed. Fig. 9 shows that the beams will begin to overlap after they have traveled a few centimeters from the face of the cluster array 60. Obviously, overlap
 10 occurs even sooner when the diodes are more closely spaced.

When tissue is stimulated with these arrays of radiation sources, a cumulative, and sometimes synergistic, effect is believed to occur, which is not seen when a single wavelength is used. It has been proposed that this effect is due, in part, to the "mixing" of photons of different wavelengths which results in three types of "two-photon events". In the first type, two different adjacent molecules are excited by
 15 different wavelength photons. In the second type, two different parts of the same molecule are excited by different wavelength photons. Both of these types of events produce excited states that would not be possible if the same molecules were stimulated with photons of only one wavelength. These "new" excited states may also make the molecule(s) involved more susceptible to certain types of decay, dissipation and reaction with each other or other unexcited molecules.

In the third type of two-photon event, a single electron is simultaneously excited by two coinciding photons of different wavelengths. The high density of photons produced by devices of small emitting surface area of the present invention enhances the probability of this type of two-photon event occurring. Assuming that all of the simultaneously-presented energy is absorbed by an electron, the resultant quantum of energy delivered by two photons is equivalent to that of a photon of a much smaller wavelength. For
 25 example, for an 880 nm photon:

$$E_{880} = \frac{h \times c}{880 \text{ nm}}$$

30 For an 820 nm photon:

$$E_{820} = \frac{h \times c}{820 \text{ nm}}$$

35 Since the effective energy of both photons will be the sum of E_{880} and E_{820} :

$$E_{\text{effective}} = E_{880} + E_{820} = \frac{h \times c}{\text{effective wavelength}}$$

40 The effective wavelength is about one-half of the average of the two original wavelengths or, in this example, approximately 425 nm. In effect, the target molecule is stimulated as if it had been hit with a single 425 nm photon.

45 The third type of two-photon event is especially significant because 425 nm is a wavelength that would normally be absorbed by skin pigments and would not penetrate very deeply into the skin. By stimulating the tissue with photons at 880 nm and 820 nm the screening effect of the skin is avoided. In certain preferred and alternative embodiments diodes of four wavelengths are used, thus creating ten different two-wavelength combinations, resulting in ten effective wavelengths during the third type of two-photon event. The effective wavelengths range from 330 nm to 475 nm which approximately corresponds to the range of highest absorption by skin pigments as shown in Fig. 2.

50 As noted above, an array of radiation sources as used in the present invention may be comprised of laser, superluminous and similar light-emitting diodes. These types of diodes are all substantially monochromatic non-gaseous radiation sources. Continuous wave diodes are preferable, because they have a higher average available power than pulsed diodes. Appropriate diodes are available from several suppliers at wavelengths of 650 nm, 660 nm, 680 nm, 750 nm, 780 nm, 800 nm, 810 nm, 820 nm, 830 nm, 840 nm, 850 nm, 860 nm, 870 nm, 880 nm, 900 nm, 904 nm, 1100 nm, 1300 nm and 1500 nm. These radiation sources

are "substantially monochromatic," as that term is used throughout this specification and the appended claims, in that, in addition to emitting light or radiation at substantially one "main" wavelength, they also emit a significantly smaller amount of radiation at other wavelengths which are close, but not identical, to the "main" wavelength. A laser diode will emit at a "main" wavelength and a few peripheral wavelengths (corresponding to multiple resonances or off-axis modes) characterized by distinct narrow spikes in its wavelength spectrum. A superluminescent or other light-emitting diodes will emit at a "main" wavelength that is at the peak of a somewhat broader continuous band of wavelengths in a wavelength spectrum. By convention, the "main" wavelength is used to identify the diode (e.g., a "800 nm diode" will emit at a "main" wavelength of 800 nm and at some other peripheral wavelengths characteristic of the material from which the diode is made). It should be noted that the number of combinations of two wavelengths leading to possible two-photon events is increased dramatically by the existence of peripheral wavelengths associated with the "main" wavelength of each type of diode. The slight variations in wavelength can lead to virtually hundreds of two-wavelength combinations and, as a result, hundreds of effective wavelengths in the visible and ultraviolet spectra.

The present invention has as one objective the delivery of significant amounts of low-power radiation to deeper tissues. For this reason, the radiation sources, while low-powered, are relatively tightly-clustered. For the arrays shown in Figs. 6-8, the average power density at the plane of the array is in the range from about 10mW/cm² to about 40 mW/cm². Due to beam divergence, absorption, reflection, refraction, scattering and other similar effects, the average power density decreases with distance from the plane of the array either through air or into tissue. It should be noted, however, that due to the small radiation surface areas of some laser diodes, certain small areas near the plane of the array or at the surface of the treated tissue (when the cluster probe 60 is placed immediately adjacent the tissue) may have a power density of at least 120 mW/cm². Should it be found that higher energy densities yield enhanced biostimulative effects without undesirable side effects, even higher powered diodes and/or somewhat more densely clustered diodes can be utilized. Of course, the energy delivered to a given area of tissue is also a function of time of exposure. Accordingly, it is also useful to speak of an energy density provided per unit area, defined as the power density multiplied by the exposure time. By this formula, a sixty second exposure of tissue with a minimum power density of 120mW/cm² (which might be found to be a minimum exposure for some significant therapeutic result) could be described as of minimum treatment energy density of 7.2 Joules/cm².

In the preferred and alternative embodiments of present invention, the radiation sources can be modulated in pulses of different frequencies ranging from 2.28 Hz to 400 kHz, including 2.28 Hz, 4.56 Hz, 9.12 Hz, 16 Hz, 18.24 Hz, 36.48 Hz, 73 Hz, 146 Hz, 292 Hz, 700 Hz, 1000 Hz, 5 kHz and 300 kHz. This is done by means of the oscillator frequency control 47 mentioned above. Other frequencies could obviously be selected. As mentioned above, the beam oscillator 45 and its frequency control 47 can also provide pulse duration modulation for continuous wave radiation sources. Thus, for the same frequency a higher average power can be obtained according to the formula: $P \text{ (average)} = P \text{ (peak)} \times \text{Pulse duration} \times \text{Frequency}$.

2. Method of Treatment

The present invention also includes a method for treatment of tissue. The method involves exposing the tissue to a plurality of radiation sources of different wavelengths. More generally, the method of treatment of the present invention involves the simultaneous exposure of the tissue to at least three different wavelengths of radiation. Any embodiment of the device of the present invention, including but not limited to those previously described, can be used to perform this method of treatment. The array of radiation sources is preferably placed directly adjacent to or on the skin such that the plane of the radiation sources is close to or comes in contact with the outermost layer of skin. Because oils and other substances on the surface of the skin may cause absorption, refraction, reflection and/or diffraction of some wavelengths of radiation and thereby decrease radiation penetration, for effectiveness these should be removed before treatment.

Devices of the present invention have been used to treat various conditions in clinical settings. The results of those clinical applications as reported by several medical doctors and physiotherapists in the United Kingdom are summarized in the following examples.

Example 1

A patient had a thirteen year history of extreme pain in the right big toe after engaging in sports activities. Upon examination, the patient was found to have a congenital Hallux Valgus or chronic "bunion". The sore toe was treated with a multi-diode biostimulation device of the present invention (660 nm, 820nm 880 nm, 950 nm) for a period of ten minutes. The patient experienced immediate relief of all pain and was able to engage in sports activities the day after treatment.

Example 2

A patient experienced post plaster pain for two weeks following an operation for Hallux Valgus on the left foot. Examination revealed pitting edema of the forefoot, slight swelling of the ankle and limited movement of the foot. The patient was treated daily for five minutes with a multi-diode biostimulation device of the present invention (660 nm, 820 nm, 880 nm, 950 nm). Treatment continued for five days. Swelling decreased after each treatment. After the third treatment, the foot was pain free, but still slightly swollen. Upon completion of treatment, all symptoms had been relieved.

Example 3

A 24 year-old patient experienced pain after surgery to repair a ruptured left anterior cruciate ligament. The patient was treated twice daily with a multi-diode biostimulation device of the present invention (660 nm, 820 nm, 880 nm, 950 nm) for a period of 4 minutes, 30 seconds. After treatment, the patient was pain free and showed signs of good tissue repair.

Example 4

A 38 year-old patient had experienced chronic fibrositis in the neck for 18 years. The patient was treated daily for two days with a 21 diode biostimulation device of the present invention (660 nm, 820 nm, 880 nm, 950 pm) for a period of 4 minutes, 30 seconds. Each treatment was followed by treatment by treatment with a 15 mW 850 nm single diode probe for two minutes. After treatment, the patient experienced 98% pain relief.

Example 5

A 35 year-old patient had experienced muscle spasms in the lower back. The patient was treated twice in a single day with a multi-diode biostimulation device of the present invention (660 nm, 820 nm, 880 nm, 950 nm) for a period of 4 minutes, 30 seconds. After treatment, the patient experienced total pain relief.

Example 6

A 33 year-old patient had experienced pain in the head of his fibula secondary to a knee injury in which the patient sustained a torn medial meniscus and ruptured anterior cruciate. The meniscus had been removed. The pain radiated down the leg and was associated with limitation of knee flexion. The patient was treated six times with a 660 nm single diode probe for a period of 4 minutes followed by treatment with a multi-diode biostimulation device of the present invention (660 nm, 820 nm, 880 nm, 950 nm) for a period of 2 minutes. Laser treatment was used in conjunction with mobilizations. After treatment, the patient experienced relief of virtually all pain and regained full range of knee flexion.

Example 7

A 21 year-old patient had experienced an inflamed gleuteal bursa. The patient was treated three times with a diode biostimulation device of the present invention (660 nm, 820 nm, 880 nm, 950 nm). After treatment, the patient showed no further pain and no reoccurrence.

Example 8

A 35 year-old patient had experienced a non-healing skin ulcer for 3 years following a motorcycle accident. The patient was treated 10 times with a multi-diode biostimulation device of the present invention (660 nm, 820 nm, 880 nm, 950 nm). After treatment, the tissue showed increased granulation (+ +) and the ulcer had decreased in size and looked healthier.

Example 9

A 70 year-old patient had experienced a diabetic ulcer following an above-knee amputation. The lateral third of the ulcer was treated with a 830 nm single diode probe and a multi-diode biostimulation device of the present invention (660 nm, 820 nm, 880 nm, 950 nm) for a period of 90 seconds each. After treatment, increased healing, granulation and de-sloughing were observed.

Preliminary indications are that the method and apparatus of the invention may be used widely for therapeutic purposes, for example, to treat inflammations, wounds, burns, chronic ulcerations including diabetic ulcers, deficient circulation, pain, nerve degeneration, eczema, shingles, infection, scars, acne, bone fractures, muscle and ligament injuries, arthritis, osteo-arthritis, rheumatoid arthritis, skin grafts, gingival irritation, oral ulcers, dental pain and swelling, cellulitis, stretch marks, skin tone, alopecia areata, trigeminal neuralgia, herpes, zoster, sciatica, cervical erosions and other conditions.

From the foregoing, it will be obvious to those skilled in the art that various modifications in the above-described method and apparatus can be made without departing from the spirit and scope of the invention. Accordingly, the invention may be embodied in other specific forms without departing from the spirit or essential characteristics thereof. Present embodiments, therefore, are to be considered in all respects as illustrative and not restrictive, the scope of the invention being indicated by the appended claims rather than by the foregoing description, and all changes which come within the meaning and range of equivalency of the claims are therefore intended to be embraced therein.

Because of the nature of light and the optical properties of tissue, power density from a single light source obtained at subcutaneous tissue is attenuated. However, when multiple crossing or overlapping beams of light are present, regions of added power density occur in the subcutaneous tissue which enable a therapeutic threshold to be achieved. The occurrence of these regions of added power density or "hot spots" enables sufficient energy density to be achieved within an acceptably short exposure time.

It has also been found as a result of laboratory research that certain cells, e.g. macrophages, when exposed to light at particular wavelengths, release a number of wound factor chemicals which can assist in the repair of damaged tissue. For example, light at 820 (laser diode), 660, 870 nm (super luminous diode) released factors that helped the proliferation of fibroblasts significantly above control levels in *in vitro* culture. Monochromatic light at 880 nm (superluminous diode) released factors that inhibited fibroblast proliferation, which may be significant when uncontrollable fibroblast proliferation needs to be prevented e.g. to prevent keloids scar tissue formation after burns. It has been suggested that the protein coating of cells may be involved and that different cells e.g. fibroblasts, T-cells, B-cells may each react at a different wavelength or set of wavelengths provided that there is more than a threshold energy density available and targeted to the cell. When a device according to the invention is used to irradiate tissue, cells along the path of the beams will react according to their sensitive wavelengths. Cells located at deeper strata may not respond to a single wavelength because of power density attenuation, but in the regions of the overlapping beams or "hot spots" or similar or different wavelengths the energy density threshold can be achieved and a useful therapeutic result can be obtained. The possibility of 2-photon events and the effective new wavelengths created allows more kinds of cells to be targeted including possibly neural cells.

Claims

1. A device for biostimulation of tissue comprising:
an array of substantially monochromatic radiation sources, said array comprising:
5 at least one such radiation source providing a first wavelength less than 830 nm;
at least one such radiation source providing a second wavelength greater than or equal to 830 nm and less than 900 nm; and
at least one such radiation source providing a third wavelength greater than or equal to 900 nm;
said radiation sources being arranged such that at least two radiation wavelengths among said first, second
10 and third wavelengths simultaneously pass directly or indirectly through a single point located within said tissue.
2. The device for biostimulation of tissue of Claim 1, wherein:
(a) said radiation sources are selected from the group consisting of superluminous diodes and laser diodes; and/or
15 (b) said first wavelength is selected from the group consisting of 650 nm, 660 nm, 680 nm, 750 nm, 780 nm, 800 nm, 810 nm and 820 nm;
said second wavelength is selected from the group consisting of 830 nm, 840 nm, 850 nm, 860nm, 870 nm, and 880 nm; and
said third wavelength is selected from the group consisting of 900 nm, 904 nm, 950 nm, 1100 nm, 1300 nm
20 and 1500 nm; and/or
(c) said first wavelength is selected from the group consisting of 660 nm and 820 nm;
said second wavelength is selected from the group consisting of 875 nm and 880 nm; and
said third wavelength is 950 nm.
- 25 3. The device for biostimulation of tissue of Claim 2, wherein said radiation sources are modulated at pulse frequencies selected from within the range of 2.28 Hz to 400 kHz.
4. The device for biostimulation of tissue of Claim 3, wherein said pulse frequencies are selected from the group consisting of 2.28 Hz, 4.56 Hz, 9.12 Hz, 16 Hz, 18.24 Hz, 36.48 Hz, 73 Hz, 146 Hz, 292 Hz, 700 Hz, 1000 Hz, 5 kHz and 300 kHz.
- 30 5. The device for biostimulation of tissue of Claims 2, 3 or 4, wherein said array comprises:
five 660 nm superluminous diodes;
one 820 nm laser diode;
ten 880 nm superluminous or laser diodes; and
five 950 nm superluminous diodes.
- 35 6. The device for biostimulation of tissue of Claim 5, wherein:
said 820 nm diode is positioned in the center of said array;
said 880 nm diodes are evenly positioned about the circumference of a circle of 10.5 mm radius from the center of said array; and
said 660 nm diodes and 950 nm diodes are alternately positioned and evenly spaced about the
40 circumference of a circle of 17 mm radius from the center of said array such that a radial line from the center of said array to each of said 660 nm or 950 nm diodes bisects the arc between two of said 880 nm diodes.
7. The device for biostimulation of tissue of Claim 6, wherein:
said 820 nm diode is positioned in the center of said array;
45 five of said 880 nm diodes are evenly positioned about the circumference of a circle of 10.5 mm radius from the center of said array;
said 660 nm diodes are evenly positioned about the circumference of a circle of 17 mm radius from the center of said array such that a radial line from the center of said array to each of said 660 nm diodes bisects the arc between two of said 880 nm diodes in said first set;
50 said 950 nm diodes are evenly positioned about the circumference of a circle of 17 mm radius from the center of said array such that a radial line from the center of said array to each of said 950 nm diodes passes through the center of one of said 880 nm diodes in said first set; and
the remaining five of said 880 nm diodes are evenly positioned about the circumference of a circle of 27 mm radius from the center of said array such that a radial line from the center of said array to each of said
55 880 nm diodes passes through the center of one of said 660 nm diodes.

8. The device for biostimulation of tissue of Claim 2, wherein said array comprises:

ten 660 nm superluminous diodes;
 one 820 nm laser diode;
 ten 880 nm superluminous or laser diodes; and
 5 ten 950 nm superluminous diodes.

9. The device for biostimulation of tissue of Claim 8, wherein:

said 820 nm diode is positioned in the center of said array;
 said 880 nm diodes are evenly positioned about the circumference of a circle of 9.5 mm radius from the center of said array;

10 said 660 nm diodes are evenly positioned about the circumference of a circle of 18 mm radius from the center of said array such that a radial line from the center of said array to each of said 660 nm diodes bisects the arc between two of said 880 nm diodes; and

said 950 nm diodes are evenly positioned about the circumference of a circle of 27 mm radius from the center of said array such that a radial line from the center of said array to each of said 950 nm diodes passes through the center of one of said 880 nm diodes.

10. A device for biostimulation of tissue comprising:

an array of substantially monochromatic radiation sources, said array comprising:

at least one such radiation source providing a first wavelength less than 830 nm;

at least one such radiation source providing a second wavelength greater than or equal to 830 nm and less than 875 nm;

at least one such radiation source providing a third wavelength greater than or equal to 875 nm and less than 900 nm; and

at least one such radiation source providing a fourth wavelength greater than or equal to 900 nm;

said radiation sources being arranged such that at least two radiation wavelengths among said first, second, third and fourth wavelengths simultaneously pass directly or indirectly through a single point located within said tissue.

11. The device for biostimulation of tissue of Claim 10, wherein said radiation sources are selected from the group consisting of superluminous diodes and laser diodes.

12. The device for biostimulation of tissue of Claim 11, wherein:

30 said first wavelength is selected from the group consisting of 650 nm, 660 nm, 680 nm, 750 nm, 780 nm, 800 nm, 810 nm and 820 nm;

said second wavelength is selected from the group consisting of 830 nm, 840 nm, 850 nm, 860 nm and 870 nm;

said third wavelength is 880 nm; and

35 said fourth wavelength is selected from the group consisting of 900 nm, 904 nm, 950 nm, 1100 nm, 1300 nm and 1500 nm.

13. A device for biostimulation of tissue comprising:

an array of substantially monochromatic light-emitting diodes, said array comprising:

at least one diode providing a first wavelength less than 800 nm; and

40 at least one diode providing a second wavelength greater than or equal to 800 nm;

said diodes being arranged such that radiation of said first and second wavelengths simultaneously passes directly or indirectly through a single point located within said tissue.

14. The device for biostimulation of tissue of Claim 13, wherein:

45 said first wavelength is selected from the group consisting of 650 nm, 660 nm, 680 nm, 750 nm and 780 nm; and

said second wavelength is selected from the group consisting of 800 nm, 810 nm, 820 nm, 830 nm, 840 nm, 850 nm, 860 nm, 870 nm, 880 nm, 900 nm, 904 nm, 950 nm, 1100 nm, 1300 nm and 1500 nm.

15. A device for biostimulation of tissue comprising:

50 an array of substantially monochromatic light-emitting diodes, said diodes being arranged in a single plane and emitting an average power density at said plane of at least 10 mW/cm².

16. A device for biostimulation of tissue comprising:

an array of substantially monochromatic light-emitting diodes, said diodes being arranged so as to provide a power density of at least 120 mW/cm² at some point on the surface of said tissue.

17. A device for biostimulation of tissue comprising:

55 an array of substantially monochromatic non-gaseous continuous wave radiation sources, said radiation sources being arranged in a single plane and emitting an average power density at said plane of at least 10 mW/cm².

18. A device for biostimulation of tissue comprising:
an array of substantially monochromatic non-gaseous continuous wave radiation sources, said radiation sources being arranged so as to provide a power density of at least 120 mW/cm² at some point on the surface of said tissue.

5 19. A system for biostimulation of tissue comprising:
a biostimulation device comprising:
an array of substantially monochromatic radiation sources, said array comprising:
at least one such radiation source providing a first wavelength less than 830 nm;
at least one such radiation source providing a second wavelength greater than or equal to 830 nm and less
10 than 900 nm; and
at least one such radiation source providing a third wavelength greater than or equal to 900 nm;
said radiation sources being arranged such that at least two radiation wavelengths among said first, second
and third wavelengths simultaneously pass directly or indirectly through a single point located within said
tissue; and
15 means in communication with said array for providing power to said radiation sources such that each of said
radiation sources emits radiation at its characteristic wavelength.

20. The system for biostimulation of tissue of Claim 2, wherein said system further comprises:
(a) means in communication with said power-providing means for modulating the output of said
sources with pulses and varying the modulation pulse frequency of said radiation sources; and/or
20 (b) means in communication with said power-providing means for varying the pulse duration of said
radiation sources; and/or
(c) means in communication with said power-providing means for timing the period that power is
provided to said radiation sources; and/or
(d) means in communication with said power-providing means for measuring the electrical conductiv-
25 ity of said tissue; and/or
(e) means in communication with said array for measuring the optical power emitted by said radiation
sources; and/or
(f) means in communication with said control panel for detecting radiation from said radiation sources;
and
30 indicator means connected to said detecting means for indicating that radiation is being emitted from said
radiation sources.

21. A system for biostimulation of tissue comprising:
a biostimulation probe comprising:
35 an array of substantially monochromatic radiation sources, said array comprising:
at least one such radiation source providing a first wavelength less than 830 nm;
at least one such radiation source providing a second wavelength greater than or equal to 830 nm and less
than 900 nm; and
at least one such radiation source providing a third wavelength greater than or equal to 900 nm;
40 said radiation sources being arranged such that at least two radiation wavelengths among said first, second
and third wavelengths simultaneously pass directly or indirectly through a single point located within said
tissue;
means in communication with said array for providing power to said radiation sources such that each of said
radiation sources emits radiation at its characteristic wavelength;
45 means in communication with said power-providing means for modulating the pulse frequency of said
radiation sources;
means in communication with said power-providing means for varying the pulse duration of said radiation
sources;
means in communication with said power-providing means for timing the period that power is provided to
50 said radiation sources;
means in communication with said array for measuring the electrical conductivity of said tissue;
means in communication with said power-providing means for measuring the optical power emitted by said
radiation sources; and
means in communication with said control panel for detecting emissions from said radiation sources.

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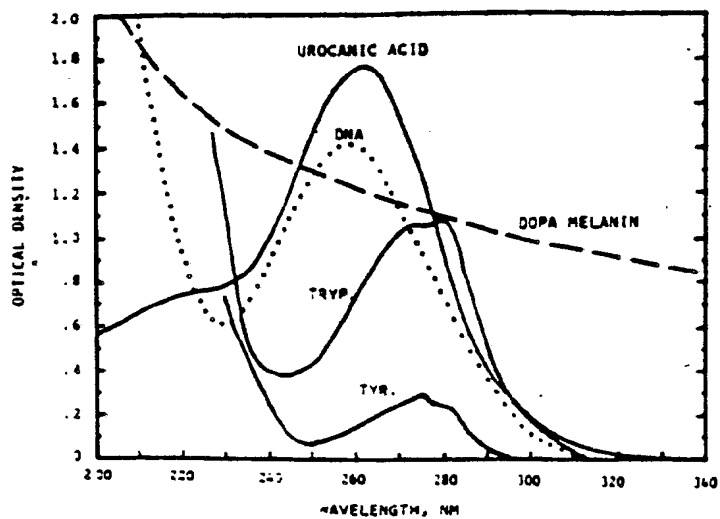


FIG. 1

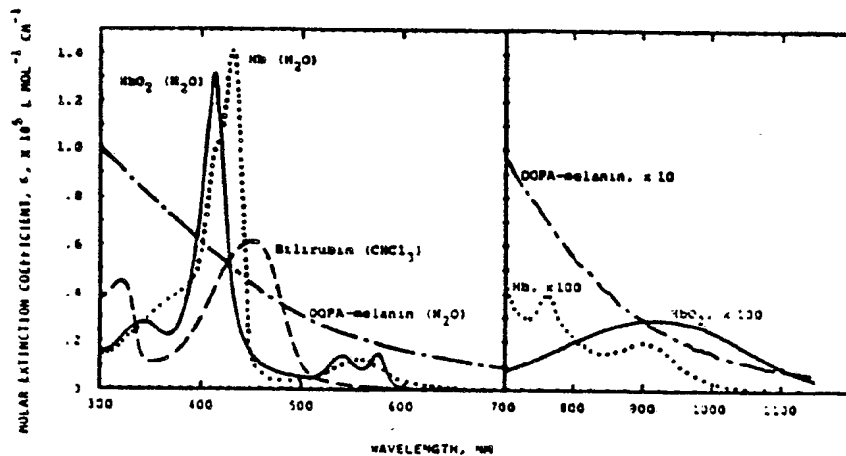


FIG. 2

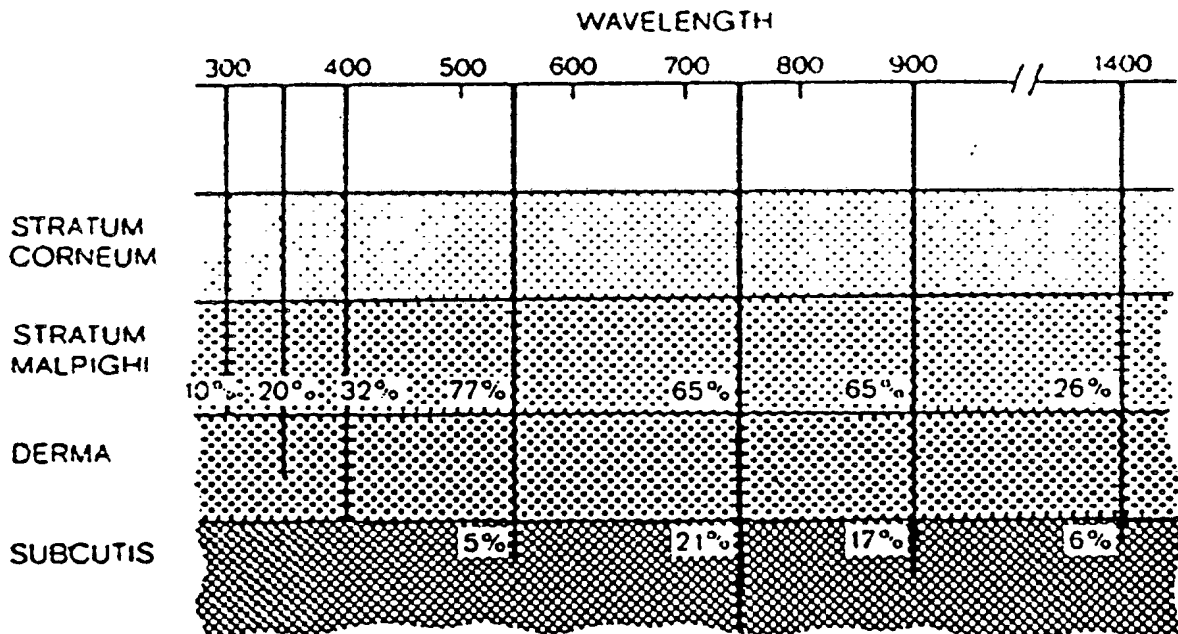
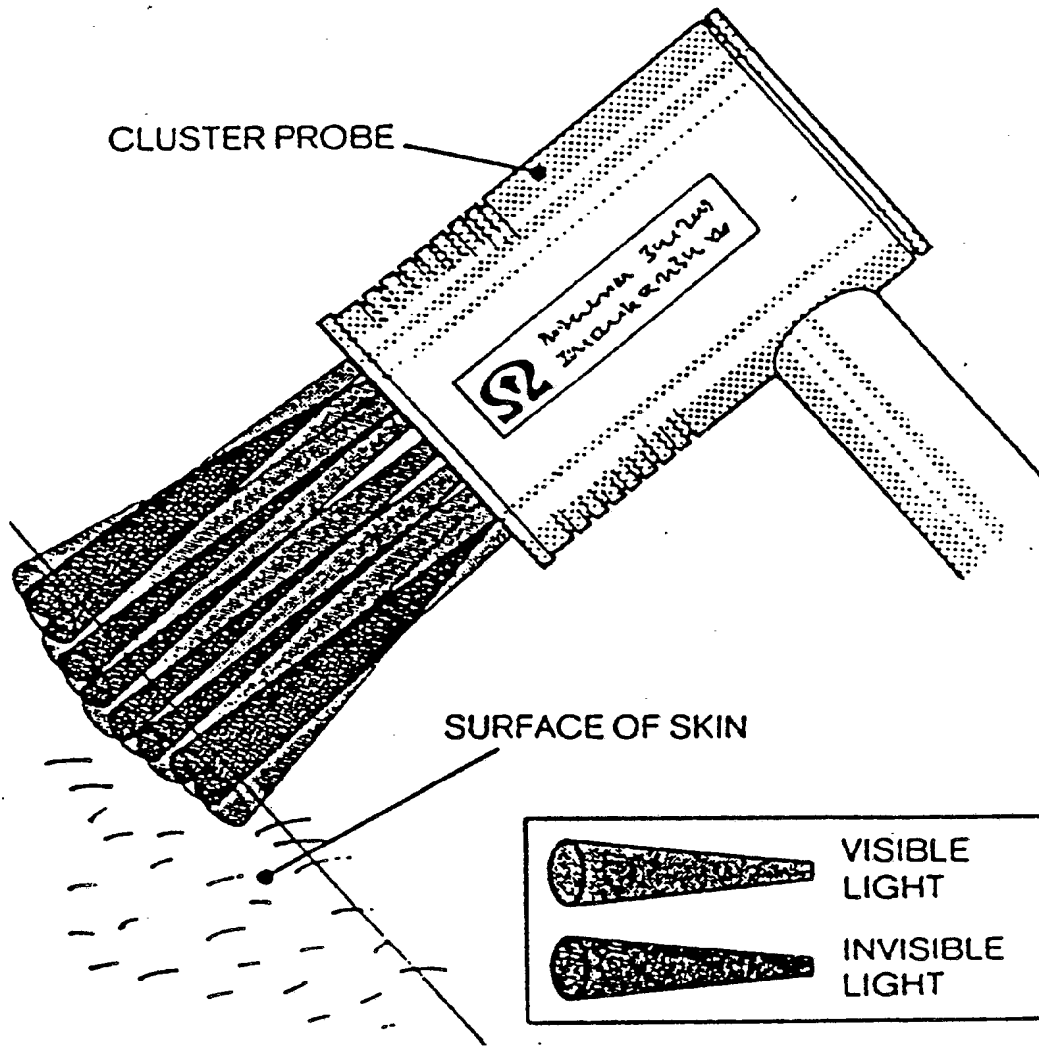


FIG. 3

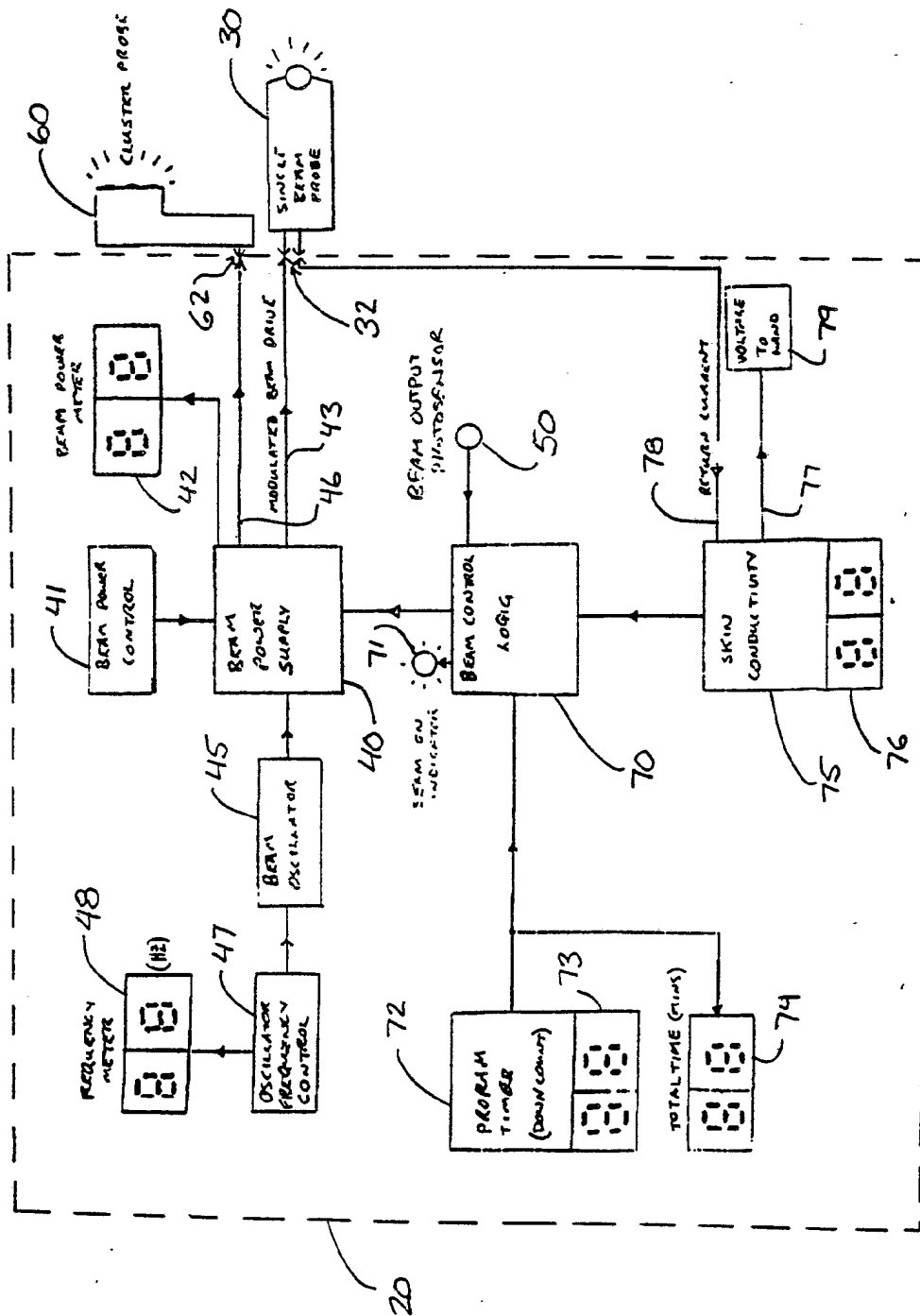


FIG. 4

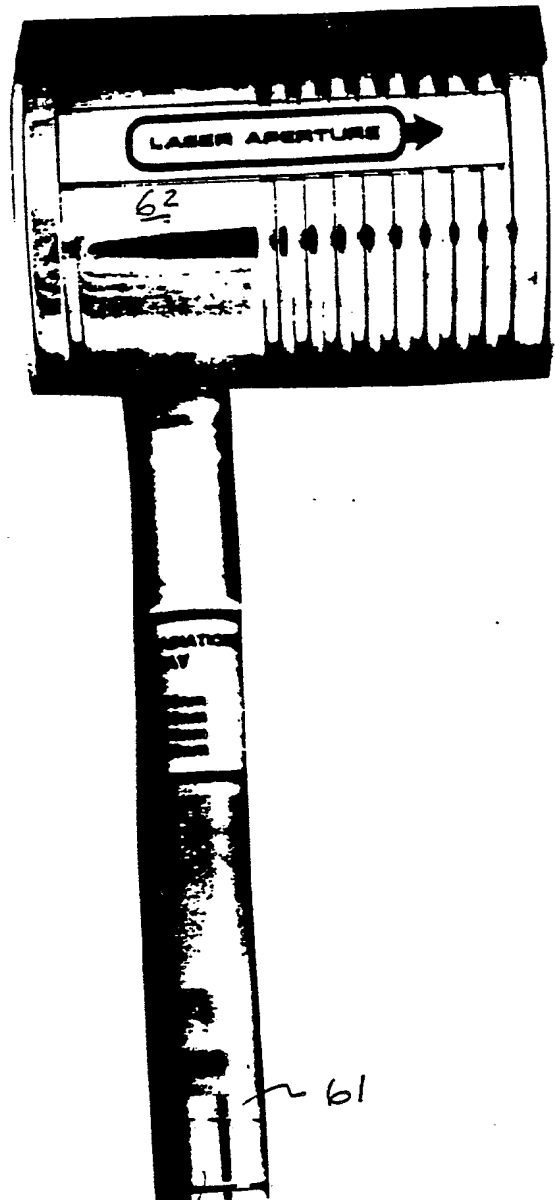


FIG. 5

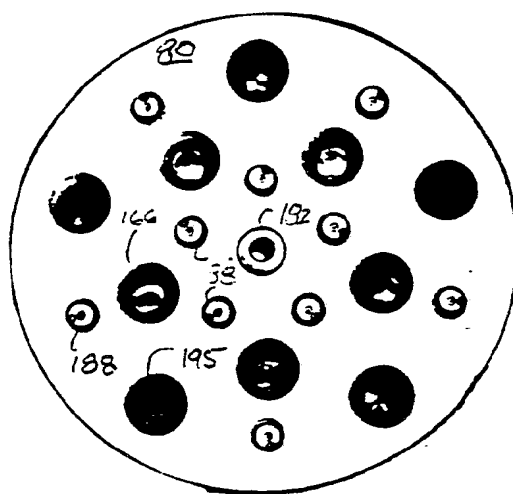


FIG. 6

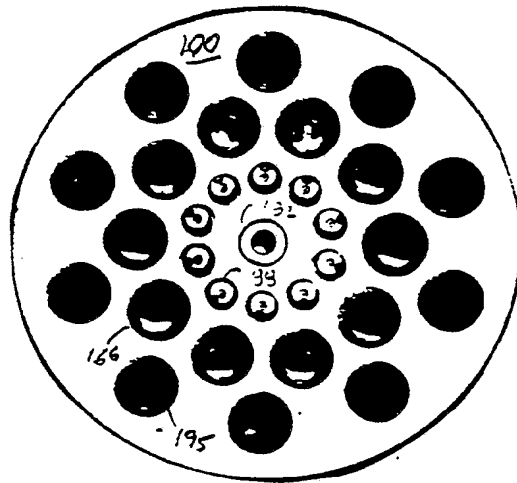


FIG. 8

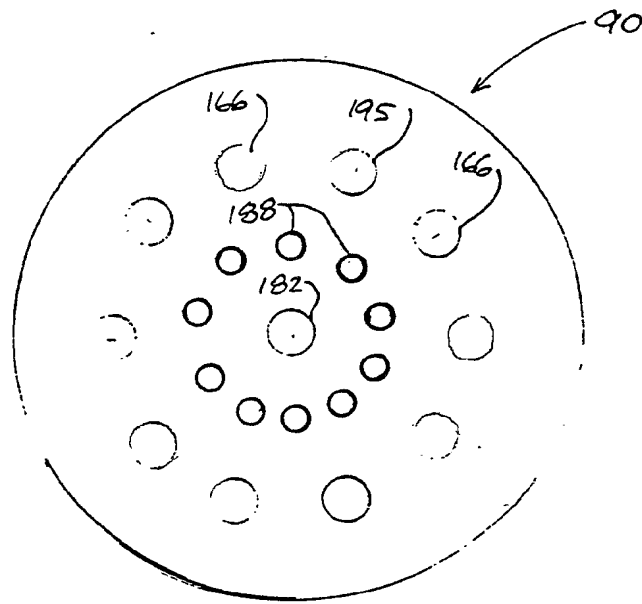


FIG. 7

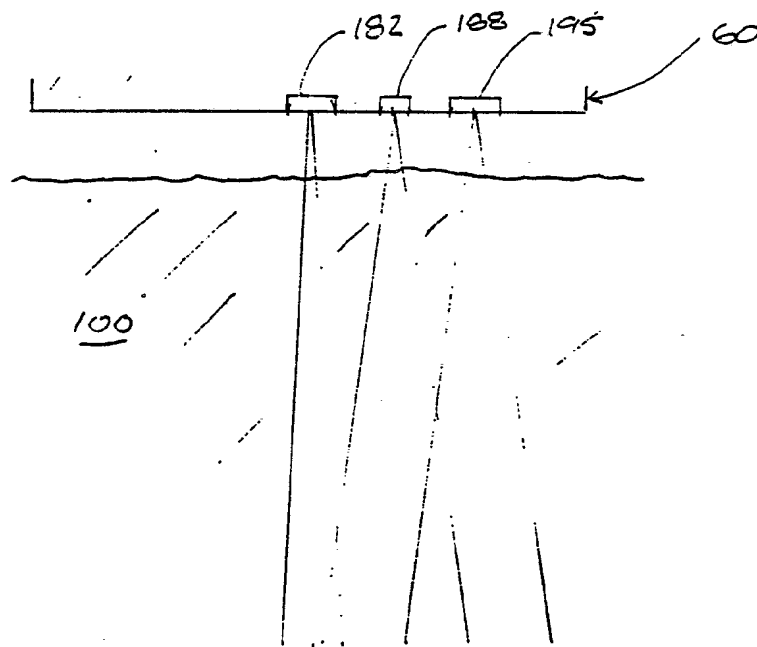


FIG. 9



DOCUMENTS CONSIDERED TO BE RELEVANT			EP 88304792.0
Category	Citation of document with indication, where appropriate, of relevant passages	Relevant to claim	CLASSIFICATION OF THE APPLICATION (Int. Cl.4)
A	CH - A5 - 651 477 (RAUNA) * Totality *	1,10, 13,15- 19,21	A 61 N 5/06
A	DE - A1 - 3 508 674 (LAMBERTZ) * Claims 7,8,10; page 4, lines 17-19; page 5, lines 19-30; page 6, lines 15-18,29; fig. 1,2 *	1,10, 13,15- 19,21	
A	FR - A2 - 2 399 256 (SKOVAJSA) * Page 2, line 30 - page 3, line 6; fig. 1-3 *	1,10, 13,15- 19,21	
A	US - A - 4 646 743 (PARRIS) * Abstract; column 2, lines 42-58; fig. 1,2 *	1,10, 13,15- 19,21	TECHNICAL FIELDS SEARCHED (Int. Cl.4)
A	DE - U - 7 622 099 (LOHER) * Claims 1-3; page 4, paragraph 4; page 5, last 5 lines *	20	A 61 N A 61 H
A	EP - A2 - 0 058 105 (JAVELLE) * Abstract; page 6, line 36 - page 7, line 12; fig. 1 *	1,10, 13,15- 19,21	
A	US - A - 4 535 784 (ROHLICEK) * Abstract; fig. 1-8 *	1	
The present search report has been drawn up for all claims			
Place of search VIENNA		Date of completion of the search 28-02-1989	Examiner NEGWER
CATEGORY OF CITED DOCUMENTS X : particularly relevant if taken alone Y : particularly relevant if combined with another document of the same category A : technological background O : non-written disclosure P : intermediate document T : theory or principle underlying the invention E : earlier patent document, but published on, or after the filing date D : document cited in the application L : document cited for other reasons & : member of the same patent family, corresponding document			



DOCUMENTS CONSIDERED TO BE RELEVANT			EP 88304792.0
Category	Citation of document with indication, where appropriate, of relevant passages	Relevant to claim	CLASSIFICATION OF THE APPLICATION (Int. Cl.4)
A	US - A - 4 112 335 (GONSER) * Column 3, lines 39-59 * --	1	
			TECHNICAL FIELDS SEARCHED (Int. Cl.4)
The present search report has been drawn up for all claims			
Place of search VIENNA		Date of completion of the search 28-02-1989	Examiner NEGWER
CATEGORY OF CITED DOCUMENTS			
X : particularly relevant if taken alone Y : particularly relevant if combined with another document of the same category A : technological background O : non-written disclosure P : intermediate document			
T : theory or principle underlying the invention E : earlier patent document, but published on, or after the filing date D : document cited in the application L : document cited for other reasons & : member of the same patent family, corresponding document			



⑫ **EUROPEAN PATENT APPLICATION**

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⑤④ **Therapeutic electromagnetic treatment.**

⑤⑦ A therapeutic treatment device is disclosed and includes a housing (12) and an incoherent light source (14) such as a flashlamp disposed in the housing. The flashlamp provides a pulsed light output for treatment of external skin disorders. To provide light to the treatment area the housing has an opening that is disposed adjacent a skin treatment area. A reflector (16) is mounted within the housing near the light source to reflect the light to the treatment area. At least one optical filter (18) and an iris (20) are mounted near the opening in the housing. Power to the lamp is provided by a pulse forming circuit that can provide a variable pulse width.

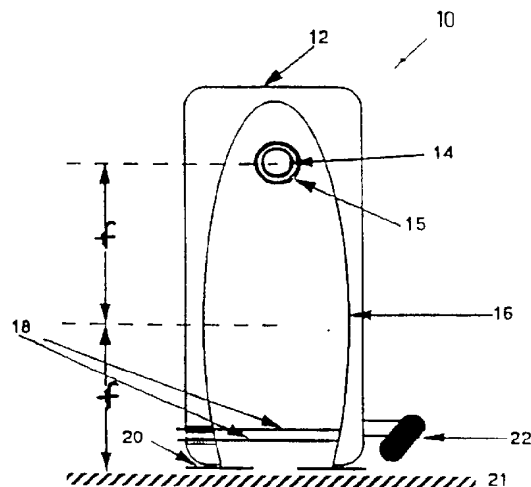


Figure 1

The present invention relates generally to the art of therapeutic electromagnetic treatment and more specifically to a method and apparatus for utilising a spatially extended pulsed light source such as a flashlamp (flash tube) for such a treatment or, efficiently focusing light from the flashlamp into optical fibres for therapeutic treatment or other applications.

It is known in the prior art to use electromagnetic radiation in medical application for therapeutic uses such as treatment of skin disorders. For example, US-A-4,298,005 (Mutzhas) describes a continuous ultraviolet lamp with cosmetic, photobiological, and photochemical applications. A treatment based on using the UV portion of the spectrum and its photochemical interaction with the skin is described. The power delivered to the skin using Mutzhas' lamp is described as 150W/m², which does not have a significant effect on skin temperature.

In addition to prior art treatment involving UV light, lasers have been used for dermatological procedures, including Argon lasers, CO₂ lasers, Nd(Yag) lasers, Copper vapor lasers, ruby lasers and dye lasers. For example, US-A-4,829,262 (Furumoto), describes a method of constructing a dye laser used in dermatology applications. Two skin conditions which may be treated by laser radiation are external skin irregularities such as local differences in the pigmentation or structure of the skin, and vascular disorders lying deeper under the skin which cause a variety of skin abnormalities including port wine stains, telangiectasias, leg veins and cherry and spider angiomas. Laser treatment of these skin disorders generally includes localised heating of the treatment area by absorption of laser radiation. Heating the skin changes or corrects the skin disorder and causes the full or partial disappearance of the skin abnormality.

Certain external disorders such as pigmented lesions can also be treated by heating the skin very fast to a high enough temperature to evaporate parts of the skin. Deeper-lying vascular disorders are more typically treated by heating the blood to a high enough temperature to cause it to coagulate. The disorder will then eventually disappear. To control the treatment depth a pulsed radiation source is often used. The depth the heat penetrates in the blood vessel is controlled by controlling the pulse width of the radiation source. The absorption and scattering coefficients of the skin also affect the heat penetration. These coefficients are a function of the constituents of skin and the wavelength of the radiation. Specifically, the absorption coefficient of light in the epidermis and dermis tends to be a slowly varying, monotonically decreasing function of wavelength. Thus, the wavelength of the light should be chosen so that the absorption coefficient is optimised for the particular skin condition and vessel size being treated.

The effectiveness of lasers for applications such as tattoo removal and removal of birth and age marks

is diminished because lasers are monochromatic. A laser of a given wavelength may be effectively used to treat a first type of skin pigmentation disorder, but, if the specific wavelength of the laser is not absorbed efficiently by skin having a second type of disorder, it will be ineffective for the second type of skin disorder. Also, lasers are usually complicated, expensive to manufacture, large for the amount of power delivered, unreliable and difficult to maintain.

The wavelength of the light also affects vascular disorder treatment because blood content in the vicinity of the vascular disorders varies, and blood content affects the absorption coefficient of the treatment area. Oxyhemoglobin is the main chromophore which controls the optical properties of blood and has strong absorption bands in the visible region. More particularly, the strongest absorption peak of oxyhemoglobin occurs at 418nm and has a band-width of 60nm. Two additional absorption peaks with lower absorption coefficients occur at 542 and 577nm. The total band-width of these two peaks is on the order of 100nm. Additionally, light in the wavelength range of 500 to 600nm is desirable for the treatment of blood vessel disorders of the skin since it is absorbed by the blood and penetrates through the skin. Longer wavelengths up to 1000nm are also effective since they can penetrate deeper into the skin, heat the surrounding tissue and, if the pulse-width is long enough, contribute to heating the blood vessel by thermal conductivity. Also, longer wavelengths are effective for treatment of larger diameter vessels because the lower absorption coefficient is compensated for by the longer path of light in the vessel.

In addition to being used for treating skin disorders, lasers have been used for invasive medical procedures such as lithotripsy and removal of blood vessel blockage. In such invasive procedures laser light is coupled to optical fibres and delivered through the fibre to the treatment area. In lithotripsy the fibre delivers light from a pulsed laser to a kidney or gallstone and the light interaction with the stone creates a shock wave which pulverises the stone. To remove blood vessel blockage the light is coupled to the blockage by the fibre and disintegrates the blockage. In either case the shortcomings of lasers discussed above with respect to laser skin treatment are present. Accordingly, a treatment device for lithotripsy and blockage removal utilising a flashlamp would be desirable.

To effectively treat an area the light from the source must be focused on the treatment area. Coupling pulsed laser light into optical fibres in medicine is quite common. The prior art describes coupling isotropic incoherent point sources such as CW lamps into small optical fibres. For example, US-A-4,757,431 (Cross, et al.) discloses a method for focusing incoherent point sources with small filaments or an arc lamp with an electrode separation of 2mm

into a small area. Point (or small) sources are relatively easy to focus without large losses in energy because of the small size of the source. Also, US-A-4,022,534 (Kishner) discloses light produced by a flash tube and the collection of only a small portion of the light emitted by the tube into an optical fibre.

However, the large dimension of an extended source such as a flashlamp make it difficult to focus large fractions of its energy into small areas. Coupling into optical fibres is even more difficult since not only must a high energy density be achieved, but the angular distribution of the light has to be such that trapping in the optical fibre can be accomplished. Thus, it is desirable to have a system for coupling the output of a high intensity, extended, pulsed light source into an optical fibre.

In order to solve the technical problems outlined above including the specificity of prior art systems and their technical complexity and expense, the device or system of the present invention is characterised by the provision of pulsed incoherent radiation.

Accordingly, in one embodiment, a wide band electromagnetic radiation source that covers the near UV and the visible portion of the spectrum would be desirable for treatment of external skin and vascular disorders. The overall range of wavelengths of the light source should be sufficient to optimise treatment for any of a number of applications. Such a therapeutic electromagnetic radiation device should also be capable of providing an optimal wavelength range within the overall range for the specific disorder being treated. The intensity of the light should be sufficient to cause the required thermal effect by raising the temperature of the treatment area to the required temperature. Also, the pulse-width should be variable over a wide enough range so as to achieve the optimal penetration depth for each application. Therefore, it is desirable to provide a light source having a wide range of wavelengths, which can be selected according to the required skin treatment, with a controlled pulse-width and a high enough energy density for application to the affected area.

Pulsed non-laser type light sources such as linear flashlamps provide these benefits. The intensity of the emitted light can be made high enough to achieve the required thermal effects. The pulse-width can be varied over a wide range so that control of thermal depth penetration can be accomplished. The typical spectrum covers the visible and ultraviolet range and the optical bands most effective for specific applications can be selected, or enhanced using fluorescent materials. Moreover, non-laser type light sources such as flashlamps are much simpler and easier to manufacture than lasers, are significantly less expensive for the same output power and have the potential of being more efficient and more reliable. They have a wide spectral range that can be optimised for a variety of specific skin treatment applications. These

sources also have a pulse length that can be varied over a wide range which is critical for the different types of skin treatments.

The scope of the invention is defined in the claims and the embodiments outlined below are specific combinations suitable for implementing the invention.

According to a first embodiment of the invention a therapeutic treatment device comprises a housing and an incoherent light source, suitably a flashlamp, operable to provide a pulsed light output for treatment, disposed in the housing. The housing has an opening and is suitable for being disposed adjacent a skin treatment area. A reflector is mounted within the housing proximate the light source, and at least one optical filter is mounted proximate the opening in the housing. An iris is mounted coextensively with the opening. Power to the lamp is provided by a variable pulse width pulse forming circuit. Thus, the treatment device provides controlled density, filtered, pulsed light output through an opening in the housing to a skin area for treatment.

According to a second embodiment of the invention a method of treatment with light energy comprises the steps of providing a high power, pulsed light output from a non-laser, incoherent light source and directing the pulsed light output to a treatment area. The pulse width of the light output is controlled and focused so that the power density of the light is controlled. Also, the light is filtered to control the spectrum of the light.

According to a third embodiment of the invention a coupler comprises an incoherent light source such as a toroidal flashlamp. A reflector is disposed around the incoherent light source and at least one optical fibre or light guide. The fibre has an end disposed within the reflector. This end collects the light from the circular lamp. In a similar coupling configuration fibres may be provided, along with a linear to circular fibre transfer unit disposed to receive light from the light source and provide light to the optical fibres. The reflector has an elliptical cross-section in a plane parallel to the axis of the linear flash tube, and the linear flash tube is located at one focus of the ellipse while the linear to circular transfer unit is located at the other focus of the ellipse.

For a better understanding of the invention, reference is made to the accompanying diagrammatic drawings, in which:

Figure 1 is a cross-sectional view of an incoherent, pulsed light source skin treatment device;

Figure 2 is a side view of the light source of Figure 1;

Figure 3 is a schematic diagram of a pulse forming network with a variable pulse width for use with the skin treatment device of Figures 1 and 2; Figure 4 is a cross-sectional view of a coupler for coupling light from a toroidal flash tube into an

optical fibre with a conical edge;

Figure 5 is a side view of a toroidal flash tube;

Figure 6 is a top view of a toroidal flash tube;

Figure 7 shows the geometry for coupling into a conical section;

Figure 8 is a cross-sectional view of a coupler for coupling light from a toroidal flash tube into an optical fibre with a flat edge;

Figure 9 is a front sectional view of a coupler for coupling light from a linear flash tube into a circular fibre bundle;

Figure 10 is a side sectional view of the coupler of Figure 9;

Figure 11 is a front view of a coupler for coupling light from a linear flash tube into an optical fibre; and

Figure 12 is a front view of a coupler for coupling light from a linear flash tube into a doped optical fibre.

In the various figures, like reference numerals are used to describe like components.

Before explaining at least one embodiment of the invention in detail it is to be understood that the invention is not limited in its application to the details of construction and the arrangement of the components set forth in the following description or illustrated in the drawings. The invention is capable of other embodiments or of being practised or carried out in various ways. Also, it is to be understood that the phraseology and terminology employed herein is for the purpose of description and should not be regarded as limiting.

Referring now to Figures 1 and 2, cross-sectional and side views of an incoherent, pulsed light source skin treatment device 10 constructed and operated in accordance with the principles of the present invention are shown. The device 10 may be seen to include a housing 12, having an opening therein, a handle 13 (Figure 2 only), a light source 14 having an outer glass tube 15, an elliptical reflector 16, a set of optical filters 18, an iris 20 and a detector 22 (Figure 1 only). Light source 14, which is mounted in housing 12, may be a typical incoherent light source such as a gas filled linear flashlamp Model No. L5568 available from ILC. The spectrum of light emitted by gas filled linear flashlamp 14 depends on current density, type of glass envelope material and gas mixture used in the tube. For large current densities (e.g., 3000 A/cm² or more) the spectrum is similar to a black body radiation spectrum. Typically, most of the energy is emitted in the 300 to 1000nm wavelength range.

To treat a skin (or visible) disorder a required light density on the skin must be delivered. This light density can be achieved with the focusing arrangement shown in Figures 1 and 2. Figure 1 shows a cross-section view of reflector 16, also mounted in housing 12. As shown in Figure 1, the cross-section of reflector 16 in a plane is perpendicular to the axis of flashlamp 14 is an ellipse. Linear flashlamp 14 is located

at one focus of the ellipse and reflector 16 is positioned in such a way that the treatment area of skin 21 is located at the other focus. The arrangement shown is similar to focusing arrangements used with lasers and efficiently couples light from flashlamp 14 to the skin. This arrangement should not, however, be considered limiting. Elliptical reflector 16 may be a metallic reflector, typically polished aluminum which is an easily machinable reflector and has a very high reflectivity in the visible, and the UV range of the spectrum can be used. Other bare or coated metals can also be used for this purpose.

Optical and neutral density filters 18 are mounted in housing 12 near the treatment area and may be moved into the beam or out of the beam to control the spectrum and intensity of the light. Typically, 50 to 100nm band-width filters, as well as low cut-off filters in the visible and ultraviolet portions of the spectrum, are used. In some procedures it is desirable to use most of the spectrum, with only the UV portion being cut off. In other applications, mainly for deeper penetration, it is preferable to use narrower band-widths. The band-width filters and the cut-off filters are readily available commercially.

Glass tube 15 is located coaxially with flashlamp 14 and has fluorescent material deposited on it. Glass tube 15 will typically be used for treatment of coagulation of blood vessels to optimise the energy efficiency of device 10. The fluorescent material can be chosen to absorb the UV portion of the spectrum of flashlamp 14 and generate light in the 500 to 650nm range that is optimised for absorption in the blood. Similar materials are coated on the inner walls of commercial fluorescent lamps. A typical material used to generate "warm" white light in fluorescent lamps has a conversion efficiency of 80%, has a peak emission wavelength of 570nm and has a bandwidth of 70nm and is useful for absorption in blood. The few millisecond decay time of these phosphors is consistent with long pulses that are required for the treatment of blood vessels.

Other shapes or configurations of flashlamp 14 such as circular, helical, short arc and multiple linear flashlamps may be used. Reflector 16 may have other designs such as parabolic or circular reflectors. The light source can also be used without a reflector and the required energy and power density may be achieved by locating light source 14 in close proximity to the treatment area.

Iris 20 is mounted in housing 12 between optical filters 18 and the treatment area and controls the length and the width of the exposed area, i.e. by collimating the output of flashlamp 14. The length of flashlamp 14 controls the maximum length that can be exposed. Typically a 8cm long (arc length) tube will be used and only the central 5cm of the tube is exposed. Using the central 5cm assures a high degree of uniformity of energy density in the exposed skin

area. Thus, in this embodiment the iris 20 (also called a collimator) will enable exposure of skin areas of a maximum length of 5cm. The iris 20 may be closed to provide a minimum exposure length of one millimetre. Similarly, the width of the exposed skin area can be controlled in the range of 1 to 5mm for a 5mm wide flashlamp. Larger exposed areas can be easily achieved by using longer flash tubes or multiple tubes, and smaller exposure areas are obtainable with an iris that more completely collimates the beam. The present invention provides a larger exposure area compared to prior art lasers or point sources and is very effective in the coagulation of blood vessels since blood flow interruption over a longer section of the vessel is more effective in coagulating it. The larger area exposed simultaneously also reduces the required procedure time.

Detector 22 (Figure 1) is mounted outside housing 12 and monitors the light reflected from the skin. Detector 22 combined with optical filters 18 and neutral density filters can be used to achieve a quick estimate of the spectral reflection and absorption coefficients of the skin. This may be carried out at a low energy density level prior to the application of the main treatment pulse. Measurement of the optical properties of the skin prior to the application of the main pulse is useful to determine optimal treatment conditions. As stated above, the wide spectrum of the light emitted from the non-laser type source enables investigation of the skin over a wide spectral range and choice of optimal treatment wavelengths.

In an alternative embodiment, detector 22 or a second detector system may be used for real-time temperature measurement of the skin during its exposure to the pulsed light source. This is useful for skin thermolysis applications with long pulses in which light is absorbed in the epidermis and dermis. When the external portion of the epidermis reaches too high a temperature, permanent scarring of the skin may result. Thus, the temperature of the skin should be measured. This can be realised using infrared emission of the heated skin, to prevent over-exposure.

A typical real-time detector system would measure the infra-red emission of the skin at two specific wavelengths by using two detectors and filters. The ratio between the signals of the two detectors can be used to estimate the instantaneous skin temperature. The operation of the pulsed light source can be stopped if a preselected skin temperature is reached. This measurement is relatively easy since the temperature threshold for pulsed heating that may cause skin scarring is on the order of 50°C or more, which is easily measurable using infrared emission.

The depth of heat penetration depends on the light absorption and scattering in the different layers of the skin and the thermal properties of the skin. Another important parameter is pulse-width. For a

pulsed light source, the energy of which is absorbed in an infinitesimally thin layer, the depth of heat penetration (d) by thermal conductivity during the pulse can be written as shown in Equation 1:

$$(Eq. 1) \quad d = 4 [k\Delta t/C\rho]^{1/2}$$

where

k = heat conductivity of the material being illuminated;

Δt = the pulse-width of the light pulse;

C = the heat capacity of the material;

ρ = density of the material.

It is clear from Equation 1 that the depth of heat penetration can be controlled by the pulse-width of the light source.

Thus, a variation of pulse-width in the range of 10^{-5} sec to 10^{-1} sec will result in a variation in the thermal penetration by a factor of 100.

Accordingly, the flashlamp 14 provides a pulse width of from 10^{-5} sec to 10^{-1} sec. For treatment of vascular disorders in which coagulation of blood vessels in the skin is the objective the pulse length is chosen to uniformly heat as much of the entire thickness of the vessel as possible to achieve efficient coagulation. Typical blood vessels that need to be treated in the skin have thicknesses in the range of 0.5mm. Thus, the optimal pulse-width, taking into account the thermal properties of blood, is on the order of 100msec. If shorter pulses are used, heat will still be conducted through the blood to cause coagulation, however, the instantaneous temperature of part of the blood in the vessel and surrounding tissue will be higher than the temperature required for coagulation and may cause unwanted damage.

For treatment of external skin disorders in which evaporation of the skin is the objective, a very short pulse-width is used to provide for very shallow thermal penetration of the skin. For example, a 10^{-6} sec pulse will penetrate (by thermal conductivity) a depth of the order of only 5 microns into the skin. Thus, only a thin layer of skin is heated, and a very high, instantaneous temperature is obtained so that the external mark on the skin is evaporated.

Figure 3 shows a variable pulse-width pulse forming circuit comprised of a plurality of individual pulse forming networks (PFN's) that create the variation in pulse-widths of flashlamp 14. The light pulse full width at half maximum (FWHM) of a flashlamp driven by a single element PFN with capacitance C and inductance L is approximately equal to:

$$(Eq.2) \quad \Delta t \approx 2[LC]^{1/2}$$

Flashlamp 14 may be driven by three different PFN's, as shown in Figure 3. The relay contacts R1', R2' and R3' are used to select among three capacitors C1, C2 and C3 that are charged by the high voltage power supply. Relays R1, R2 and R3 are used to select the PFN that will be connected to flashlamp 14. The high voltage switches S1, S2 and S3 are used to discharge the energy stored in the capacitor of the

PFN into flashlamp 14. In one embodiment L1, L2 and L3 have values of 100mH, 1mH and 5mH, respectively, and C1, C2 and C3 have values of 100mF, 1mF and 10mF, respectively.

In addition to the possibility of firing each PFN separately, which generates the basic variability in pulse-width, additional variation can be achieved by firing PFN's sequentially. If, for example, two PFN's having pulse-width Δt_1 and Δt_2 are fired, so that the second PFN is fired after the first pulse has decayed to half of its amplitude, then an effective light pulse-width of this operation of the system will be given by the relation: $\Delta t \approx \Delta t_1 + \Delta t_2$.

The charging power supply typically has a voltage range of 500V to 5kV. The relays should therefore be high voltage relays that can isolate these voltages reliably. The switches S are capable of carrying the current of flashlamp 14 and to isolate the reverse high voltage generated if the PFNs are sequentially fired. Solid-state switches, vacuum switches or gas switches can be used for this purpose.

A simmer power supply (not shown in Figure 3) may be used to keep the flashlamp in a low current conducting mode. Other configurations can be used to achieve pulse-width variation, such as the use of a single PFN and a crowbar switch, or use of a switch with closing and opening capabilities.

Typically, for operation of flashlamp 14 with an electrical pulse-width of 1 to 10msec, a linear electrical energy density input of 100 to 300J/cm can be used. An energy density of 30 to 100J/cm² can be achieved on the skin for a typical flashlamp bore diameter of 5mm. The use of a 500 to 650nm bandwidth transmits 20% of the incident energy. Thus, energy densities on the skin of 6 to 20J/cm² are achieved. The incorporation of the fluorescent material will further extend the output radiation in the desired range, enabling the same exposure of the skin with a lower energy input into flashlamp 14.

Pulsed laser skin treatment shows that energy densities in the range of 0.5 to 10J/cm² with pulse-widths in the range of 0.5msec are generally effective for treating vascular related skin disorders. This range of parameters falls in the range of operation of pulsed non-laser type light sources such as the linear flashlamp. A few steps of neutral density glass filters 18 can also be used to control the energy density on the skin.

For external disorders a typical pulse-width of 5 microsecond is used. A 20J/cm electrical energy density input into a 5mm bore flashlamp results in an energy density on the skin of 10J/cm². Cutting off the hard UV portion of the spectrum results in 90% energy transmission, or skin exposure to an energy density of close to 10 J/cm². This energy density is high enough to evaporate external marks on the skin.

Device 10 can be provided as two units: a lightweight unit held by a physician using handle 13, with

the hand-held unit containing flashlamp 14, filters 18 and iris 20 that together control the spectrum and the size of the exposed area and the detectors that measure the reflectivity and the instantaneous skin temperature. The power supply, the PFN's and the electrical controls are contained in a separate box (not shown) that is connected to the hand-held unit via a flexible cable. This enables ease of operation and easy access to the areas of the skin that need to be treated.

The invention has thus far been described in conjunction with skin treatment. However, using a flashlamp rather than a laser in invasive treatments provides advantages as well. Procedures such as lithotripsy or removal of blood vessel blockage may be performed with a flashlamp. Such a device may be similar to that shown in Figures 1 and 2, and may use the electronics of Figure 3 to produce the flash. However, to properly couple the light to an optical fibre a number of couplers 40, 80 and 90 are shown in Figures 4 and 8-10, respectively.

Coupler 40 includes an optical source of high intensity incoherent and isotropic pulsed light such as a linear flash tube 42, a light reflector 44 which delivers the light energy to an optical fibre 46. The latter has a generally conical edge in the embodiment of Figure 4. Optical fibre 46 transfers the light from light collection system 44 to the treatment area. In general, coupler 40 couples pulsed light from a flash tube into an optical fibre and has applications in medical, industrial and domestic areas.

For example, coupler 40 may be used in material processing to rapidly heat or ablate a portion of a material being processed, or to induce a photo-chemical process. Alternatively, coupler 40 may be used in a photography application to provide a flash for picture taking. Using such a coupler would allow the flash bulb to be located inside the camera, with the light transmitted to outside the camera using an optical fibre. As one skilled in the art should recognise coupler 40 allows the use of incoherent light in many applications that coherent or incoherent light has been used in the past.

To provide for coupling the light to an optical fibre, flash tube 42 has a toroidal shape, shown in Figures 5 and 6, and is disposed inside reflector 44. In addition to the toroidal shape other shapes, such as a continuous helix, may be used for flash tube 42. However, a helical tube is more difficult to manufacture than a toroidal tube. Referring now to Figure 6, flash tube 42 is generally in the shape of a torus, but is not a perfect torus since the electrodes located at the end of the torus have to be connected to the power source. This does not create a significant disturbance in the circular shape of flash tube 42, since the connection to the electrodes can be made quite small.

Reflector 44 collects and concentrates the light, and has a cross-section of substantially an ellipse, in

a plane perpendicular to the minor axis of the toroidal flash tube 42. The major axis of this ellipse preferably forms a small angle with the major axis of toroidal lamp 42. The exact value of the angle between the ellipse axis and the main axis of lamp 42 depends on the Numerical Aperture (NA) of the optical fibre.

The toroidal flash tube is positioned so that its minor axis coincides with the focus of the ellipse. The other focus of the ellipse is at the edge of optical fibre 46. Reflector 44 may be machined from metal with the inner surfaces polished for good reflectivity. Aluminum is a very good reflector with high reflectivity in the visible and ultraviolet, and it may be used for this purpose. The reflector can be machined in one piece and then cut along a surface perpendicular to the main axis of the device. This will enable integration of the toroidal flash tube into the device.

As shown in Figure 4, the edge of optical fibre 46 is a cone with a small opening angle, so that the total area of the fibre exposed to the light from the flash tube is increased. Referring now to Figure 7 the geometry for coupling light into a conical tip is shown. It is assumed here that the light comes from a region in space with a refractive index of n_2 and that the conical section of the fibre (as well as the rest of the fibre core) has a refractive index of n_1 .

Not all the light rays hitting the cone are trapped in it. For light rays that propagate in a plane that contains the major axis of the system, a condition can be derived for the angle of a ray that will be trapped and absorbed in the fibre. This condition is shown in Equation 3.

$$\sin(\mu_{\text{crit}}) = \cos(\beta) - (n_1^2/n_2^2 - 1)^{1/2} \sin(\beta) \quad (\text{Eq. 3})$$

Light will be trapped in the conical portion of the optical fibre if the incidence angle μ is larger than μ_{crit} calculated from Equation 3. Trapping is possible only if $n_1 > n_2$. If the medium outside of the fibre is air, $n_2 = 1$. Not all of the light trapped in the conical section of the fibre will also be trapped in the straight portion of the fibre if a fibre with a core and a cladding is used. If a fibre with a core and no cladding is used (air cladding), then all the rays captured in the conical section of the fibre will also be trapped in the straight section of the fibre.

The configuration shown in Figure 4 can also be used with a fluid filling the volume between the reflector and the optical fibre. A very convenient fluid for this purpose may be water. Water is also very effective in cooling the flashlamp if high repetition rate pulses are used. The presence of a fluid reduces the losses that are associated with glass to air transitions, such as the transition between the flashlamp envelope material and air. If a fluid is used in the reflector volume, then its refractive index can be chosen such that all the rays trapped in the conical section are also trapped in the fibre, even if core/cladding fibres are used.

Another way of configuring the fibre in the reflector is by using a fibre with a flat edge. This configuration is shown in Figure 8 and has trapping efficiency very close to the trapping efficiency of the conical edge. Many other shapes of the fibre edge, such as spherical shapes, can also be used. The configuration of the fibre edge also has an effect on the distribution of the light on the exit side of the fibre and it can be chosen in accordance with the specific application of the device.

The device may be used with a variety of optical fibres. Single, or a small number of millimetre or sub-millimetre diameter fibres, will typically be used in invasive medical applications. In other applications, particularly in industrial and domestic applications, it may be preferable to use a fibre having a larger diameter, or a larger bundle of fibres, or a light guide.

Figures 9 and 10 show a coupler 90 for coupling a linear flash tube 92 through a linear to circular fibre transfer unit 94 to a fibre bundle 96. A reflector 98 has an elliptical cross-section, shown in Figure 10, in a plane parallel to the axis of linear flash tube 92 in this embodiment. Tube 92 is located on one focus of the ellipse while the linear side of linear to circular bundle converter 94 is located at the other focus of the ellipse. This configuration is relatively simple to manufacture and commercially available linear to circular converters such as 25-004-4 available from General Fibre Optics may be used. This configuration is particularly useful for larger exposure areas of the fibre, or for flash illumination purposes.

The energy and power densities that can be achieved by this invention are high enough to get the desired effects in surface treatment or medical applications. For the embodiment shown in Figure 4 the total energy and power densities can be estimated as follows. For a typical toroidal lamp with a 4mm bore diameter and a major diameter of 3.3cm an electrical linear energy density input of 10J/cm into the lamp can be used with a 5μsec pulse width. The light output of the lamp will be 5 to 6J/cm for optimal electrical operating conditions. For the reflector shown in Figure 4, 50% of the light generated in the lamp will reach the lower focus. Thus, a total energy flux on the focus of 25 to 30J may be obtained. For embodiments shown in Figure 4 or Figure 8 the total cross-section area of reflector at the focal plane has a cross-section of 0.8cm².

Energy densities on the order of 30 to 40J/cm² at the entrance to the fibre should be attained with this cross-section. This corresponds to power densities of 5 to 10MW/cm², which are the typical power densities used in medical or material processing applications.

For longer pulses, higher linear electrical energy densities into the lamp can be used. For a 1msec pulse to the flash tube a linear electrical energy density of 100J/cm can be used. The corresponding energy density at the focal area would be up to

300J/cm². Such energy densities are very effective in industrial cleaning and processing applications as well as in medical applications.

Alternative embodiments for coupling the optical fibre to an extended light source such as a linear flashlamp are shown in Figures 11 and 12. In the embodiment of Figure 11 an optical fibre 101 is wound around a lamp 102 and a lamp envelope 103. Some of the light that is produced by the light source is coupled into the fibre. If the light rays are propagating in the direction that is trapped by the fibre then this light will propagate in the fibre and it can be used at a fibre output 104. One limitation of this configuration is the fact that most of the light emitted by lamp 103 travels in a direction perpendicular to the surface of lamp 103 and cannot be trapped in fibre 101.

The embodiment shown in Figure 12 overcomes this problem. A doped optical fibre 105 is wound around lamp 102 and envelope 103, rather than an undoped fibre such as fibre 101 of Figure 11. The dopant is a fluorescent material which is excited by the radiation emanating from lamp 102 and radiates light inside the fibre. This light is radiated omnidirectionally and the part of it that is within the critical angle of fibre 105 is trapped and propagates through the fibre and can be used at fibre output 104. The angle of light that is trapped in the fibre is the critical angle of the material from which the optical fibre or optical wave guide is made. For a fibre (or optical wave guide) in air this angle is given by $\sin \alpha = 1/n$.

Typically for glass or other transparent materials $n = 1.5$ and $\alpha = 41.8^\circ$. This corresponds to a trapping efficiency of more than 10% of the light emitted by fluorescence inside the fibre. If we assume a 50% efficiency of the fluorescence process we find out that more than 5% of the light produced by the lamp is trapped and propagated down the fibre. For example, a 4" (10.2cm) lamp with a linear electrical energy input of 300J/inch (118 J/cm) and 50% electrical to light conversion efficiency would couple 2.5% of its electrical energy into the fibre. This corresponds, for the 4" (10.2cm) lamp case to a total light energy of 30J of light. This embodiment has the additional advantage of transferring the wavelength emitted by the lamp to a wavelength that may be more useful in some of the therapeutic or processing applications mentioned before. Thus, fluorescent material doped in the fibre can be chosen in accordance with an emission wavelength determined by the specific application of the device.

Thus, it should be apparent that there has been provided in accordance with the present invention a flashlamp and coupler that fully satisfy the objectives and advantages set forth above. Although the invention has been described in conjunction with specific embodiments thereof, it is evident that many alternatives, modifications and variations will be apparent to those skilled in the art. Accordingly, it is intended to

embrace all such alternatives, modifications and variations that fall within the spirit and broad scope of the appended claims.

Claims

1. A therapeutic treatment device characterised in that an incoherent light source (14) is operable to provide a pulsed light output for treatment.
2. A treatment device as claimed in claim 1 further characterised in that a variable pulse width pulse forming circuit is electrically connected to said light source.
3. A treatment device as claimed in any one of the preceding claims, further characterised in that said light source is a flashlamp (14).
4. A treatment device as claimed in any one of the preceding claims, further characterised in that said light source comprises means for providing pulses having a width in the range of between substantially 0.5 and 10 microsec and an energy density of the light on the skin of up to about 10J/cm², whereby the light treats external disorders of the skin, such as: tattoos, pigmented lesions or birth and age marks.
5. A treatment device as claimed in any one of the preceding claims, further characterised in that said light source (14) is mounted in a housing (12) suitable for being disposed adjacent a skin treatment area, said housing having a reflector (16) mounted therein proximate said light source, and said housing having an opening, with an iris (20) mounted about said opening, and at least one optical filter (18) mounted proximate said opening.
6. A treatment device as claimed in claim 5, further characterised in that a means (18) for providing controlled energy density, filtered, pulsed light output through said opening and said iris to a skin area for treatment is provided.
7. A treatment device as claimed in claim 5 or 6, further characterised in that a power supply is connected to and external of said housing, wherein said housing includes a handle (13).
8. A device as claimed in any one of the preceding claims, further characterised in that a plurality of optical fibres (96), each having an end disposed within a reflector (98) and a linear to circular fibre transfer unit (94) is disposed to receive light from the light source (92) and provide light to the opt-

ical fibres.

9. A device as claimed in claim 8, wherein a reflector (98) has an elliptical cross-section in a plane parallel to the axis of a light source which comprises a linear flash tube (92), and wherein the linear flash tube is located at one focus of the ellipse while the linear to circular transfer unit (94) is located at the other focus of the ellipse. 5 10
10. A system for providing pulsed light characterised in that:
a pulsed toroidal flash tube incoherent light source (42, 92) has a reflector (44) disposed thereabout, said reflector having a cross-section of substantially an ellipse, in a plane perpendicular to the minor axis of the toroidal flash tube; and at least one optical fibre (46) having an end disposed within said reflector. 15 20
11. A system as claimed in claim 10, further characterised in that the end of the optical fibre has a cone shape. 25
12. A system as claimed in claim 10, further characterised in that the end of the optical fibre is flat.
13. A system as claimed in any one of claims 10 to 12, further characterised in that the optical fibre is air clad. 30

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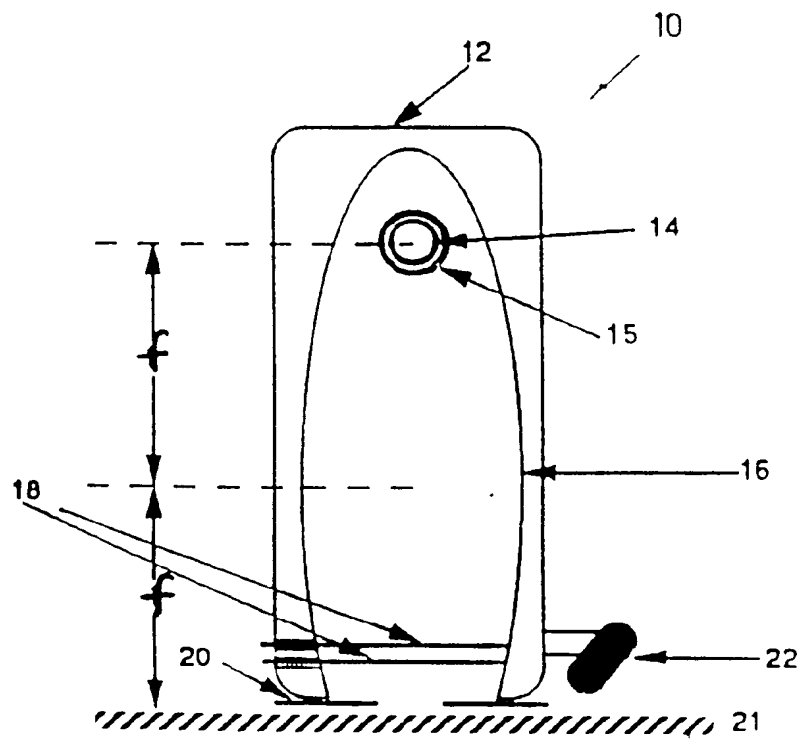


Figure 1

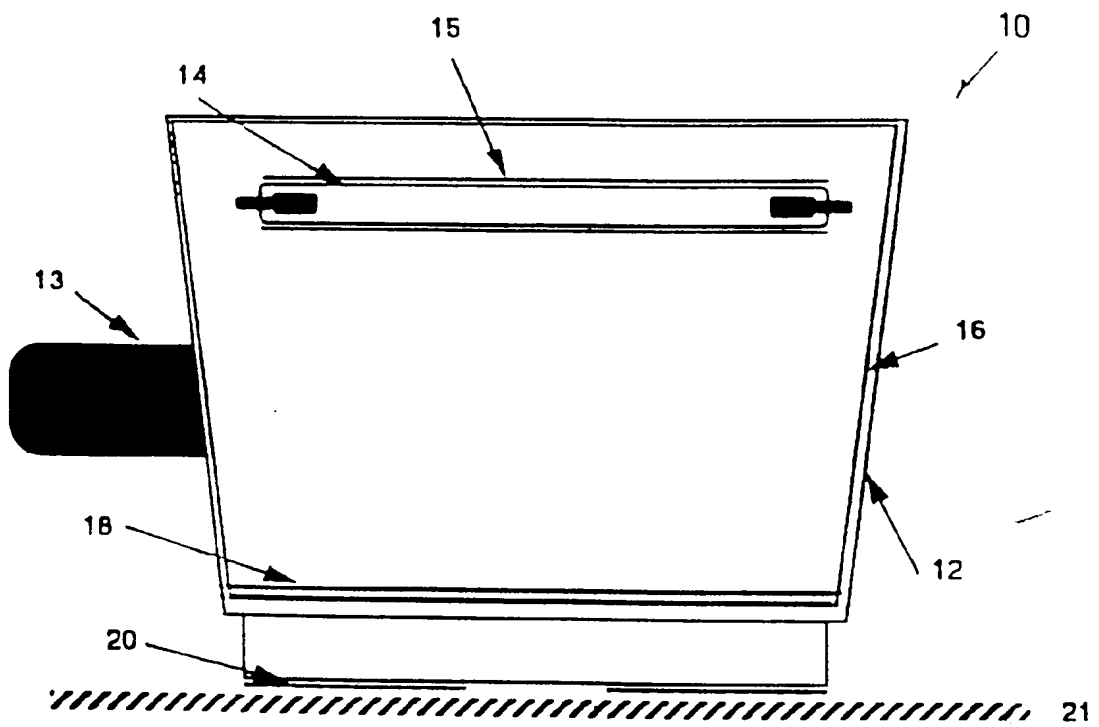


Figure 2

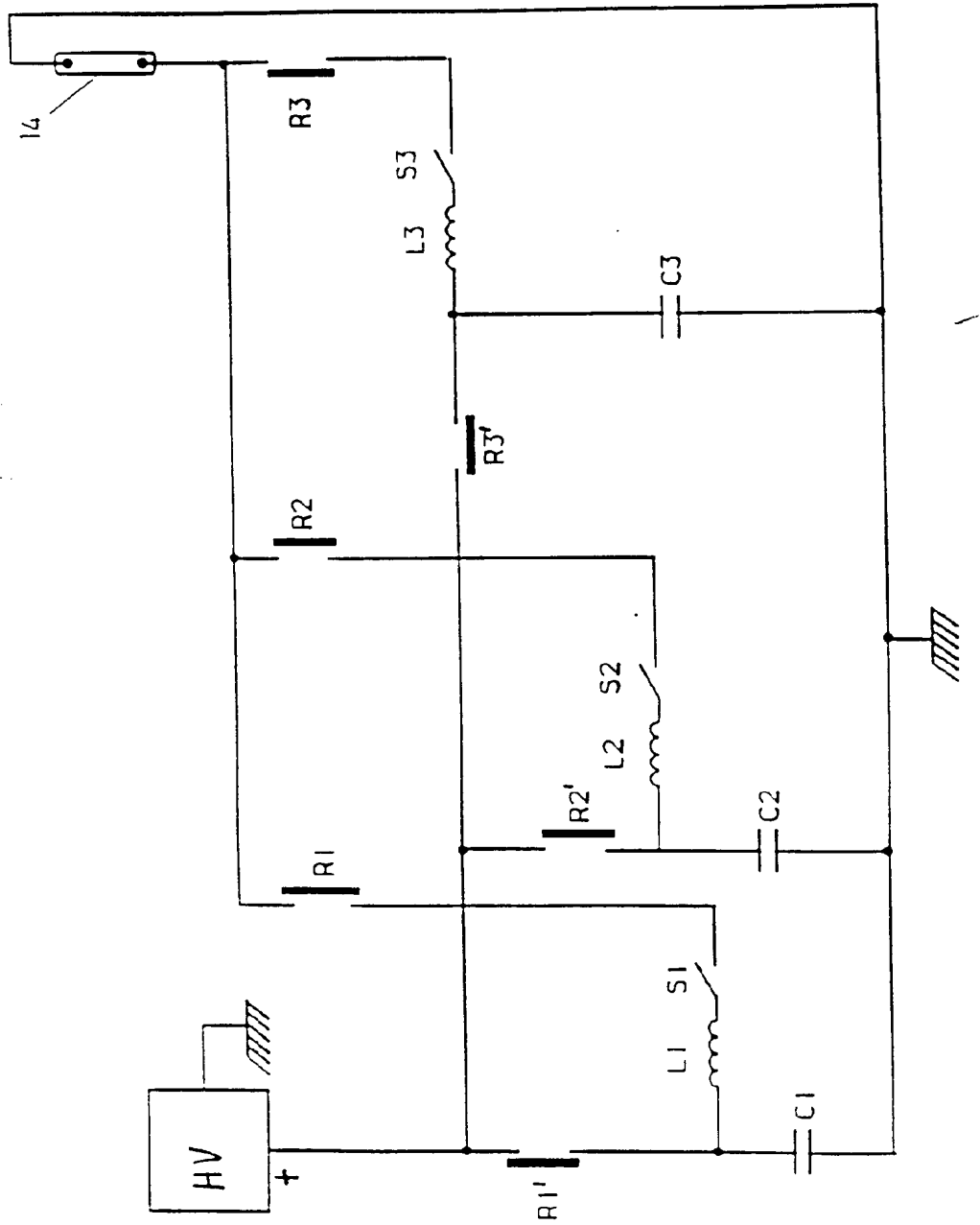


Figure 3

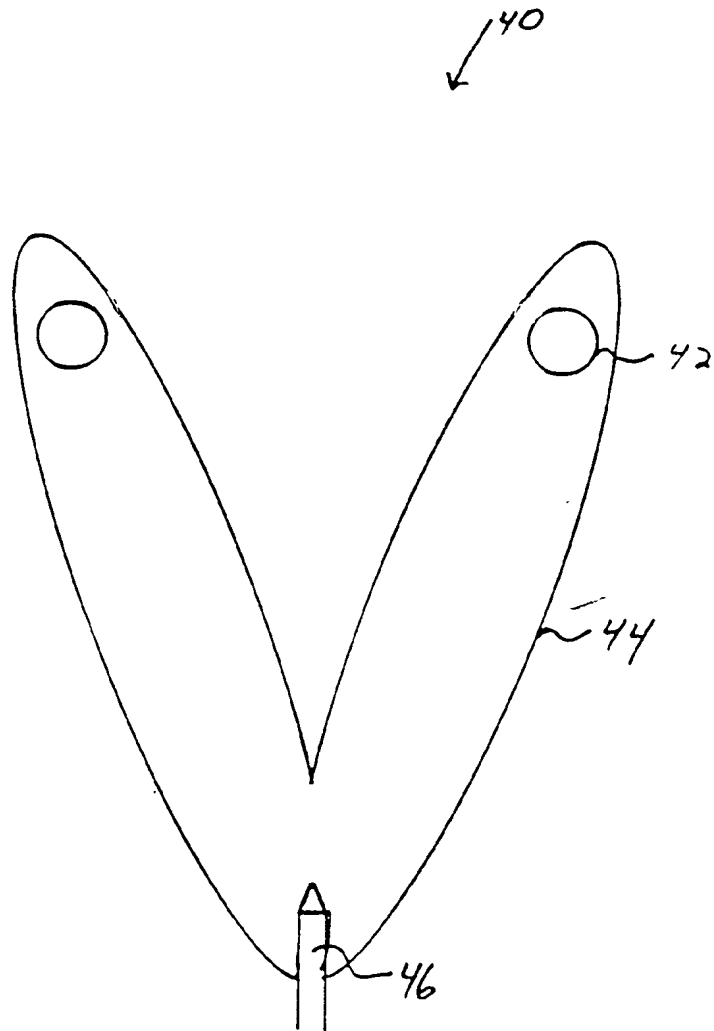


Figure 4

Figure 5

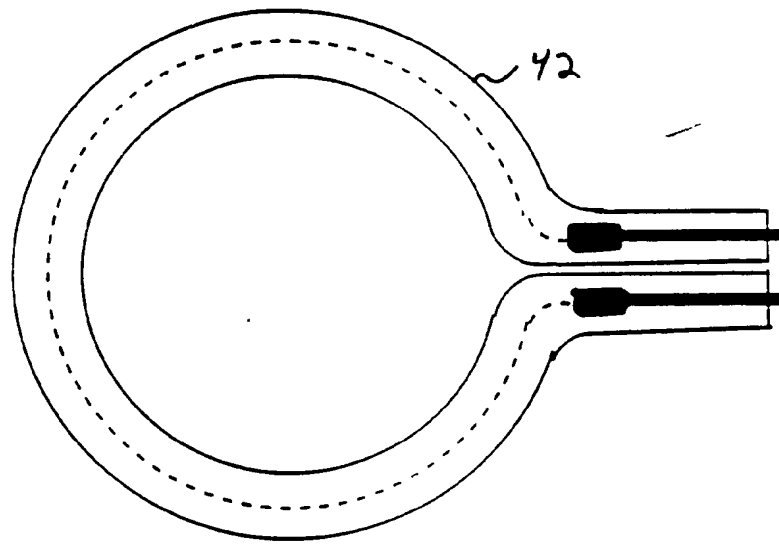
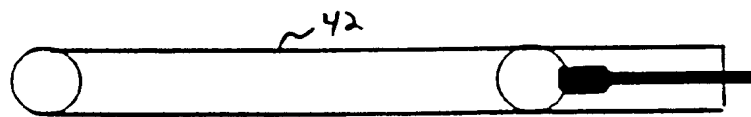


Figure 6

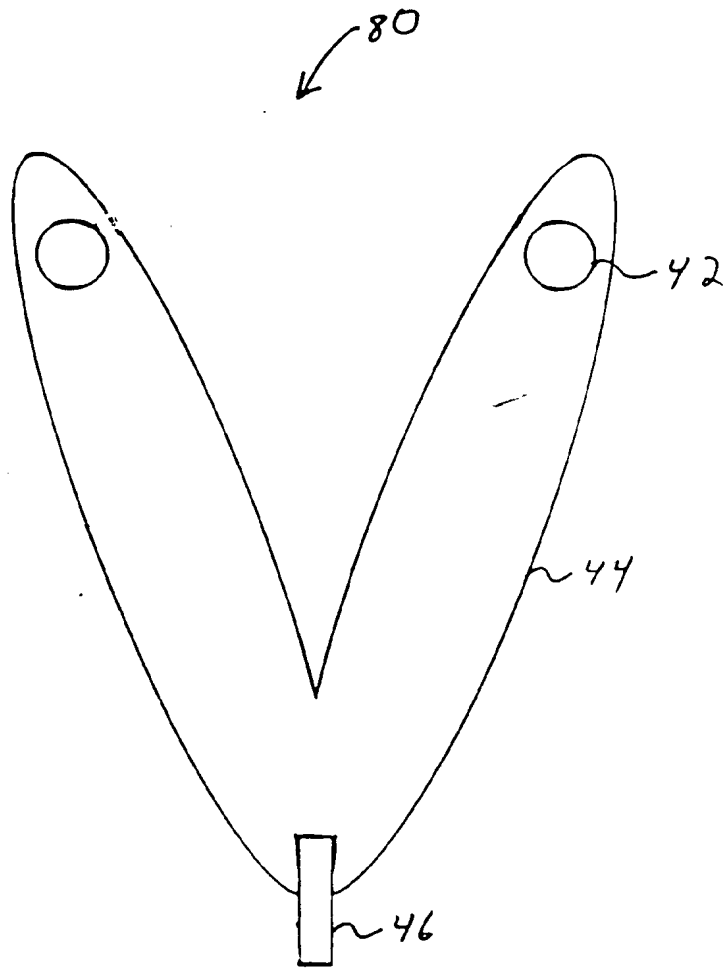


Figure 8

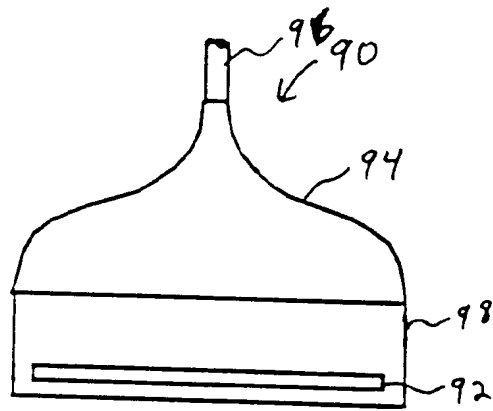


Figure 9

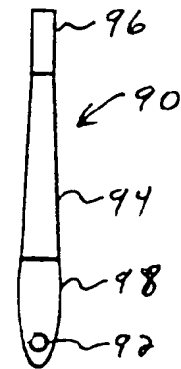


Figure 10

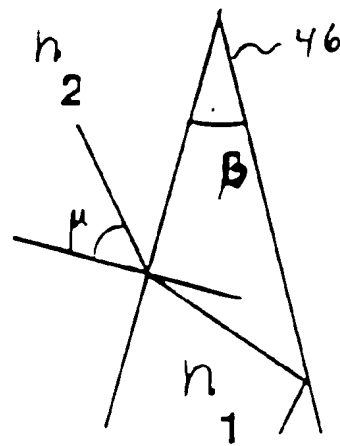


Figure 7

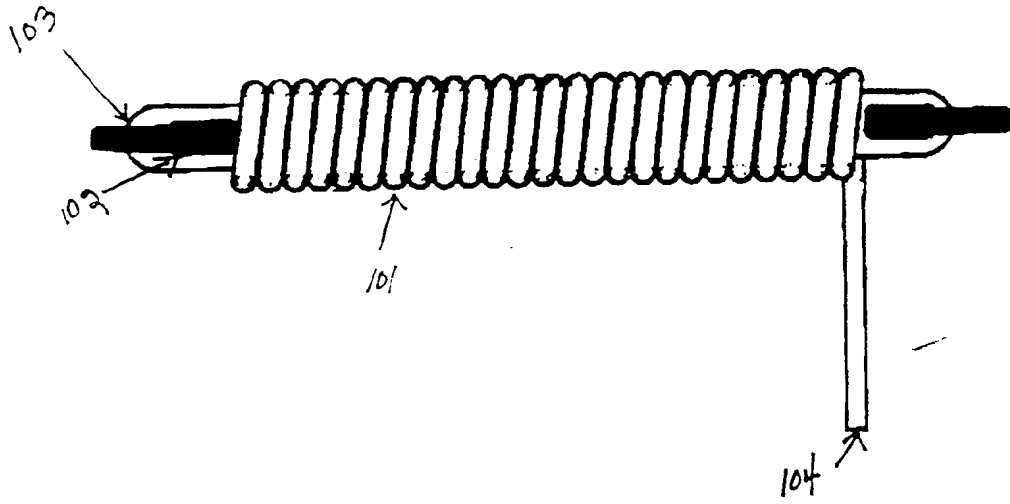


Figure 11

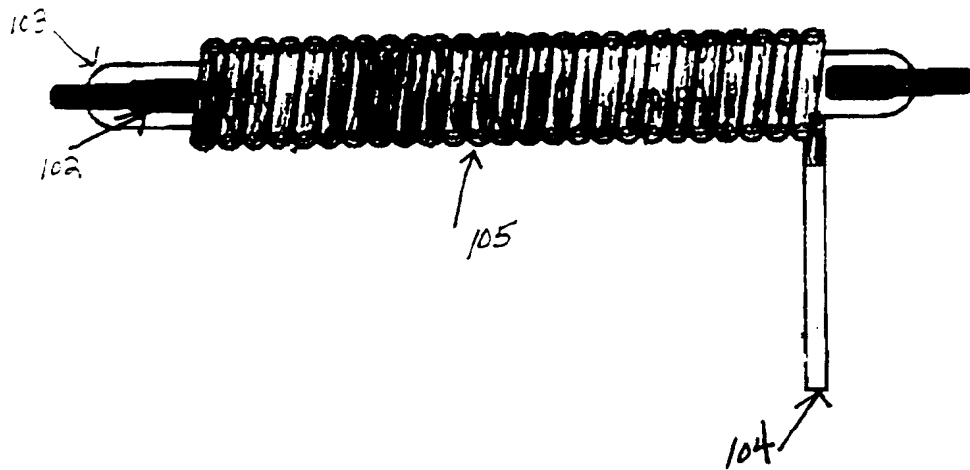


Figure 12



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(54) **Radial medical laser delivery device.**

(57) The present invention involves a medical delivery system capable of emitting radiation with wavelengths between 190 nm and 16 μ m in one or more essentially directed, predetermined patterns. It includes at least one solid optical fiber, having a core (2) and a cladding (3) on the core. The cladding has a refractive index smaller than the core, having an input end suitably configured to connect to an appropriate radiation source and having a distal end in the proximity of which two or more grooves (5-7) are penetrating into the core. The grooves have at least partial reflector capability so as to deflect radiation thereto radially in one or more predetermined patterns. The invention also includes methods of performing medical procedures utilizing the afore-said device.

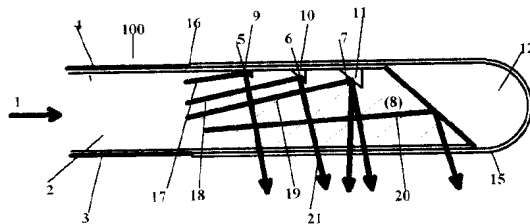


Fig 1

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BACKGROUND OF THE INVENTION

1. Field of the Invention

This invention relates to a laser delivery device, and more particularly to such delivery devices that emit radiation radially from the distal end of an optical fiber.

2. Prior Art Statement

Technological change in laser delivery devices is rapidly taking place in the laser medical field with the onset of minimally invasive procedures such as laser laparoscopy. The laparoscopist, a physician or surgeon who performs laparoscopies, is often challenged with positioning the delivery device, i.e., the optical fiber(s), at angles radially to the laparoscope axis in order to irradiate the target perpendicularly. However, in many cases moving a laparoscope radially is very difficult or is impossible. As an alternative, the laparoscope, which is normally rigid, may have an adjustable fiber deflector called a bridge. The bridge may be adjusted at the proximal end causing radial movements to the distal end of the fibers. This adjustment is, however, limited by the bend radius of the fibers and/or the bridge device and cannot offer full capabilities. Therefore, techniques to emit radiation radially from the distal end of the fiber without bending are needed.

Reflecting tips secured on the distal fiber end, such as metal caps incorporating a mirror surface at a 45° angle relative to the fiber axis are state of the art and have been used successfully in procedures such as lithotripsy with high pulse powered (Q-switched) Yttrium Aluminium Garnet Lasers.

For many surgical procedures requiring an even illumination (such as prostate treatment or photodynamic therapy) the point source-like radiation pattern from this known device is ill suited.

The state of the art devices used in photodynamic therapy incorporate a glue, i.e. epoxy, containing cap with scattering medium dispersed in it. These caps can produce a relatively homogeneous radial pattern. However, the output is diffuse and they are somewhat limited in power handling capability due to the limitations of the glue.

In summary, the present state of the art for radial laser radiation delivery is restricted to either point sources (size of the source comparable to the fiber cross section) or to essentially diffuse radiators with limited power handling capabilities. United States Patent No. 4,740,047 describes a point source type of device using a cut fiber with a reflective surface to deflect a beam for lateral application.

While methods to control the fiber tip temperature aimed at preventing damage to the distal tip of the laser delivery device have been described in United States Patent No. 5,057,099 no control method has been described to prevent or limit damage to the tissue itself that seems applicable to treatments such as laser prostatectomy. Thus, while this recently issued patent allows for temperature control to optimize particular surgical or medical procedures, it does not address or satisfactorily resolve the need for proper lateral and radial delivery of laser beams to satisfy varied needs for varied procedures.

Thus, the prior art neither teaches nor renders obvious the present invention device set forth herein.

SUMMARY OF THE INVENTION

Described is a device capable of delivering high laser power at selected angles or any angle essentially radially to the axis of an optical fiber. The fiber emits the laser radiation from a wider area at the distal end in a well directed, essentially non-diffuse pattern with a plurality of reflective surfaces, having different angles or sizes, within the fiber itself.

Surgical procedures, such as transurethral laser prostatectomy, are beneficially performed using preferred embodiments of the device. The device may comprise feedback control mechanisms from the tissue to regulate radiation delivery dosimetry with procedural requirements.

BRIEF DESCRIPTION OF THE DRAWINGS

The invention, together with further objects, advantages, aspects and features thereof, will be more clearly understood from the following description taken in connection with the accompanying drawings:

Figure 1 is a side view of a radial medical radiation delivery device using air pockets created by the core and a transparent cap for total reflection;

Figure 2 shows another radial medical radiation delivery device that can be freely positioned inside a transparent, inflatable balloon incorporating temperature sensing fibers as well, placed to irradiate the prostate;

Figure 3 is a detailed view of Figure 2 showing reflective metal coating used for deflection;

Figure 4 is a cross section of Figure 3;

Figure 5 shows a conventional state of the art Photo Dynamic Therapy Delivery device;

Figure 6 shows a delivery device with spiral grooves; and,

Figure 7 shows a power control system operated by sensing through the same fiber.

DESCRIPTION OF THE PREFERRED EMBODIMENTS

It is an object of this invention to provide a new and improved radial-laser delivery device to overcome the disadvantages of prior radial laser delivery devices, such as power handling capability, area of coverage, extent of coverage, radially directedness of radiation from an extended source, etc. By "radial" and "radially" are meant extending outwardly from the central axis of a fiber and not parallel thereto. In this application, they are meant to include extending outwardly at right angles as well as at any other angles and to include full circumference and only partial circumference radiation.

Another object of this invention is to describe a control mechanism and an improved device method to carry out treatments such as laser prostatectomy and photodynamic therapy.

Figure 1 illustrates a side view of present invention device 100, a typical preferred embodiment of the invention, at its distal end. The optical fiber 1 has a core 2, a cladding 3 and one or more protective coating layers 4. Core 2 is grooved on one side, and grooves 5, 6 and 7 are of increasing size and/or angles, as shown. Core 2 distal end 8 is encapsulated with a protective, transparent cap 15 over a predetermined length so as to cover all the grooves 5, 6 and 7; this resulting in a series of air pockets 9, 10, 11 and 12. The cap can be affixed to the fiber by any medically safe glue 16. If the inclination of the fronts of the grooves (facing incoming radiation) measured from the most inclined ray 17, 18 and 19 travelling in the fiber 1 is chosen such that it is lower than the angle of the total reflection limit between the optical fiber core and air, all rays coming through the fiber from the proximal end (input end of the radiation source, or laser) will be totally reflected and thus exit in radial direction as shown by the typical arrows such as arrow 21.

By progressively increasing the depth of each groove towards the distal end 8 of the fiber 1, more and more radiation is diverted from the axial path into the radial direction resulting in the desired extended directed radiation. This creates a defined, predetermined area of radiation application that is much greater than a reflected point source.

Figure 2 now illustrates how another such device 102 is employed to shrink the prostate gland and thus provide a free passage in the urethra. As known, the prostate gland can swell and thus result in an inconvenience for a high number of men, particularly at higher age, in as much as the ure-

thra is thus partially blocked and the free flow of urine can be obstructed. It is known that by irradiating the prostate, and thus degenerating and shrinking it this inconvenience can be removed, and a free passage restored. In order to perform this procedure in a controlled and safe manner a present invention radial medical delivery device 102 comprising an optical fiber 31, a multilumen channel 32, an inflatable balloon 33 as well as temperature sensing fibers, such as fibers 34 and 35, is introduced into the urethra 35. Fiber 31 has grooves 41, 42 and 43 and cut end 44, as shown. After inflating the balloon that is transparent to the radiation wavelength used in the procedure (example, 1064 nm) radiation is directed at the prostate 36. The inclinations of the grooves 41, 42 and 43 and cut tip 44, vary in this example, so that the radiation represented incoming by arrows 45, 46, 47 and 48, and outgoing by arrows 51, 52, 53 and 54, converges toward the prostate 36.

The radiation is thus effectively penetrating the urethra wall 38 in a less concentrated form than it is hitting the prostate, thus limiting the damage done to it.

The balloon 33 can be cooled by gas or liquid to further protect the prostate wall. By feeding the temperature reading obtained via sensing fibers 34 and 35 back to a laser power control, an optimum radiation level can be obtained.

In this example of a preferred embodiment of the radial medical delivery device, the grooves 41, 42, 43 and the cut tip 44 of the distal end 50 of the fiber, shown in part in Figure 3 are at least partially covered by a reflective metal 57, 58 and 59 (such as gold) to deflect the radiation. Dark areas 61, 62 and 63, for example, receive substantially no radiation.

Figure 4 shows a cross section and illustrates how, by flattening the lower side 60 of the fiber 31 focusing in all but the desired dimension and direction may be avoided.

The superiority over the present state of the art will now be clear: Compared to a single reflective (or totally reflective) point source on the end of a fiber the energy density penetrating through the balloon and the urethra wall is much lower and a certain degree of focusing can be achieved, if desired, towards the the present invention for prostate degeneration, a fiber of synthetic silica could be used to deliver the laser power at 1064 nm. The fiber for sensing the tissue temperature may be of silver halide semi-crystalline material (transmitting a wavelength range between 4 μ m and 16 μ m).

Any other available or known materials may be used for the fiber for a particular application without exceeding the scope of the present invention. For example, it can be equally possible to make the radial medical radiation delivery device employing

a silver halide fiber for the laser delivery itself.

In this case a CO or CO₂ laser can be used as a radiation source with wavelength of around 5 μm and typically 10.6 μm . In this case, the same fiber through which the laser radiation passes for irradiating the tissue can also be used to measure tissue temperature as well, as illustrated in Figure 6 and Figure 7.

Figure 6 shows present invention device 106 a silver halide fiber consisting of core 91 and clad 92. In this case, circular cut angled grooves 93 and 94 and tip 96, as well as a transparent cap 95 are included. While the laser radiation 116 is targeted towards the tissue 107, the temperature radiation from tissue 108 is picked up by the fiber and transmitted via a reflector 112 formed at tip 96, in the optical path of the transmission, and fed back as shown by arrow. As shown in Figure 7, this feedback is diverted via prism 114 towards a laser control module 122 thus controlling the power output of the laser 123 in line with procedural requirements.

It is evident that in some instances it may be preferable from a manufacturing standpoint to fuse a tip of a fiber containing the grooves on to another fiber, thus effectively in the end obtaining a device similar in operative characteristics to the ones described so far, and the present invention device may include a fiber formed of such joined sections without exceeding the scope of the present invention.

Clearly, in some instances it may be advantageous to build the delivery system of more than one delivery fiber processing the characteristic as described so far in this invention, for instance in order to provide higher flexibility of the device while still maintaining a certain total cross section, a fiber bundle may be used, without exceeding the scope of the present invention. Such bundles may have fibers with identical configurations but slightly staggered to enhance transmission, or may form components of a single desired configuration, depending upon the application(s) intended.

Obviously, numerous modifications and variations of the present invention are possible in light of the above teachings. It is therefore understood that within the scope of the appended claims, the invention may be practiced otherwise than as specifically described herein.

Claims

1. A medical delivery system capable of emitting radiation with wavelengths between 190 nm and 16 μm in one or more essentially directed, predetermined patterns, which comprises:
 - at least one solid optical fiber, having a core and a cladding on said core and said

cladding having a refractive index smaller than the core, having an input end suitably configured to connect to an appropriate radiation source and having a distal end in the proximity of which two or more grooves are penetrating into the core, said grooves having at least partial reflector capability so as to deflect radiation thereto radially in one or more predetermined patterns.

2. A medical delivery system as claimed in claim 1, further characterized by a cap being placed over said at least one fiber at its distal end and over said two or more grooves, and by the enclosure of gas pockets in the grooves by means of said cap.
3. A radial delivery system as claimed in claim 2, further characterized by filling the grooves with a material having a significantly lower reflective index than the fiber core.
4. A medical radiation delivery system as claimed in claim 1, wherein the grooves have a reflective coating on at least one side.
5. A medical radiation delivery system as claimed in claim 1, wherein said at least one fiber is a quartz glass or synthetic silica fiber and the radiation transmitted is between 180 and 3000nm.
6. A medical radiation delivery system as claimed in claim 1, wherein the fiber is a silver halide fiber and the radiation transmitted is between 4 μm and 16 μm . In this case the cladding on the core may be air.
7. A medical radiation delivery system as claimed in claim 1, wherein the grooves are only on one side of the device.
8. A medical radiation delivery system as claimed in claim 1, wherein the grooves have inclinations which vary in the device so as to give a radiation pattern converging at a predetermined distance from the fiber axis.
9. A medical radiation delivery system as claimed in claim 1, which further includes means for collecting through the fiber, the heat radiation from the irradiated surface, thereby controlling the energy level delivered.
10. A medical radiation delivery device system as claimed in claim 1, which further includes one or more temperature control sensors affixed on to an inflatable balloon transparent at least

over its cylindrical portion to the radiation wavelength used and incorporating the radiation delivery fiber in the inside of said inflatable balloon.

11. A medical radiation delivery device system as claimed in claim 10, wherein said fiber is located within said inflatable balloon in a movable manner.

12. A medical radiation delivery system as claimed in claim 1, further comprising dosage monitoring fibers affixed to an inflatable balloon transparent at least over an essential part of its surface to the radiation wavelength used.

13. A method of performing a laser prostatectomy procedure, comprising:

(a) the inserting of a cystoscope into the urethra

(b) positioning a device which includes at least one solid optical fiber, having a core and a cladding on said core and said cladding having a refractive index smaller than the core, having an input end suitably configured to connect to an appropriate radiation source and having a distal end in the proximity of which two or more grooves are penetrating into the core, said grooves having at least partial reflector capability so as to deflect radiation thereto radially in one or more predetermined patterns; and,

(c) irradiating the prostate area to be degenerated.

14. The method of claim 13 wherein said device is further characterized by a cap being placed over said at least one fiber at its distal end and over said two or more grooves, and by the enclosure of gas pockets in the grooves by means of said cap.

15. The method of claim 14 wherein said device is further characterized by filling the grooves with a material having a significantly lower reflective index than the fiber core.

16. A method of performing a prostate degeneration procedure comprising:

(a) inserting at least the distal end of a device into the urethra, which includes at least one solid optical fiber, having a core and a cladding on said core and said cladding having a refractive index smaller than the core, having an input end suitably configured to connect to an appropriate radiation source and having a distal end in the proximity of which two or more grooves are

penetrating into the core, said grooves having at least partial reflector capability so as to deflect radiation thereto radially in one or more predetermined patterns, and which further includes one or more temperature control sensors affixed on to an inflatable balloon transparent at least over its cylindrical portion to the radiation wavelength used and incorporating the radiation delivery fiber in the inside of said inflatable balloon;

(b) positioning it as necessary;

(c) inflating the balloon; and,

(d) irradiating the prostate area to be degenerated.

17. The method of claim 16, wherein said fiber is located within said inflatable balloon in a movable manner.

18. Method of performing photodynamic therapy, comprising:

(a) applying a photosensitive substance to the area to be treated or to the distal end of the device set forth below;

(b) inserting a device which includes at least one solid optical fiber, having a core and a cladding on said core and said cladding having a refractive index smaller than the core, having an input end suitably configured to connect to an appropriate radiation source and having a distal end in the proximity of which two or more grooves are penetrating into the core, said grooves having at least partial reflector capability so as to deflect radiation thereto radially in one or more predetermined patterns; and,

(c) irradiating the tissue to the intended dosage level.

19. The method of claim 18, further characterized by a cap being placed over said at least one fiber at its distal end and over said two or more grooves, and by the enclosure of gas pockets in the grooves by means of said cap.

20. The method of claim 18, further characterized by filling the grooves with a material having a significantly lower reflective index than the fiber core.

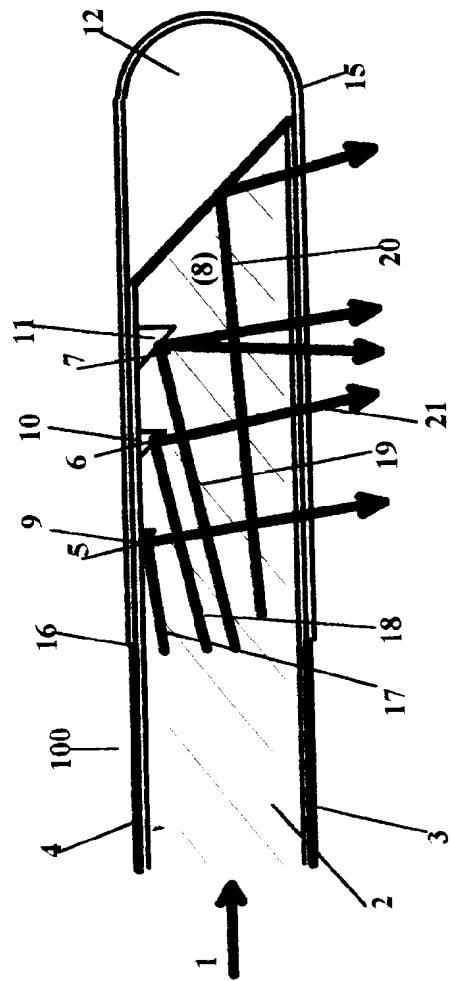


Fig 1

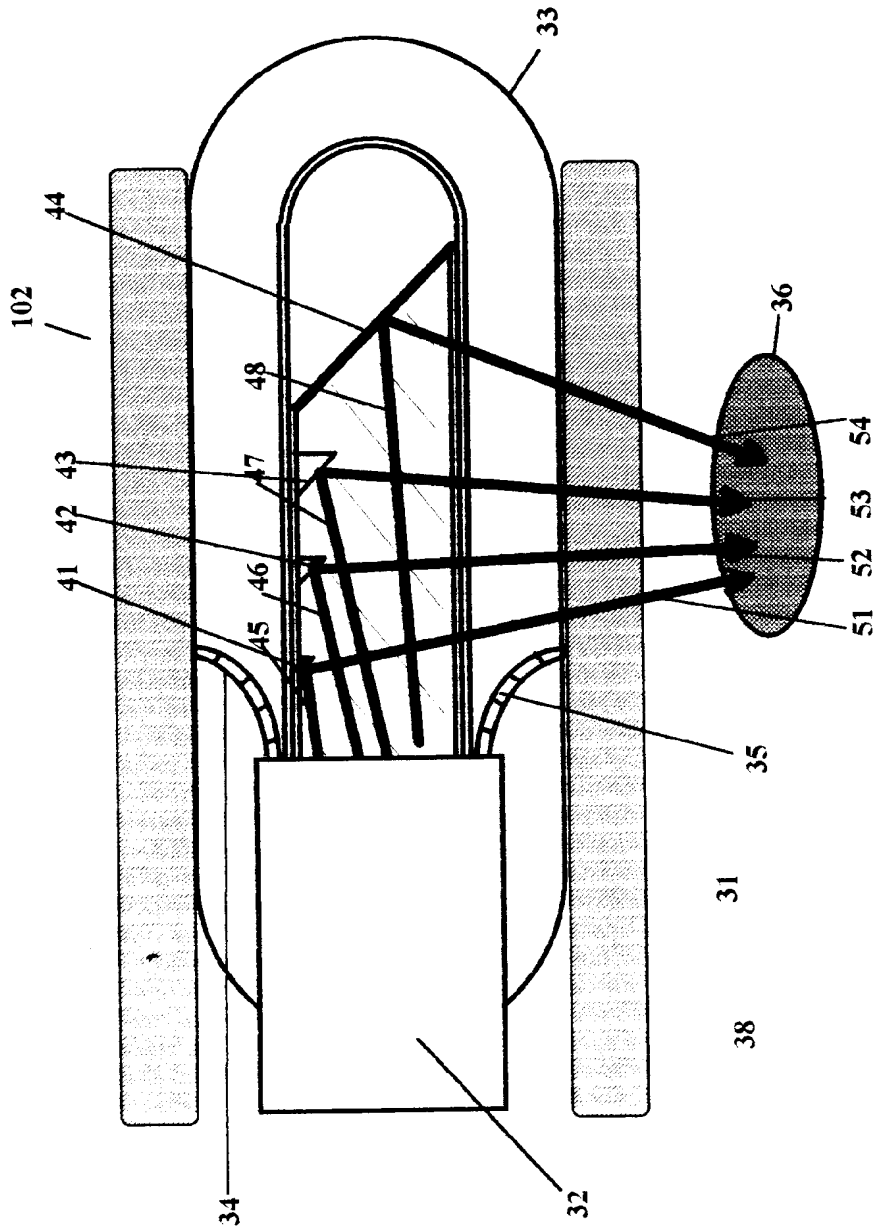


Fig 2

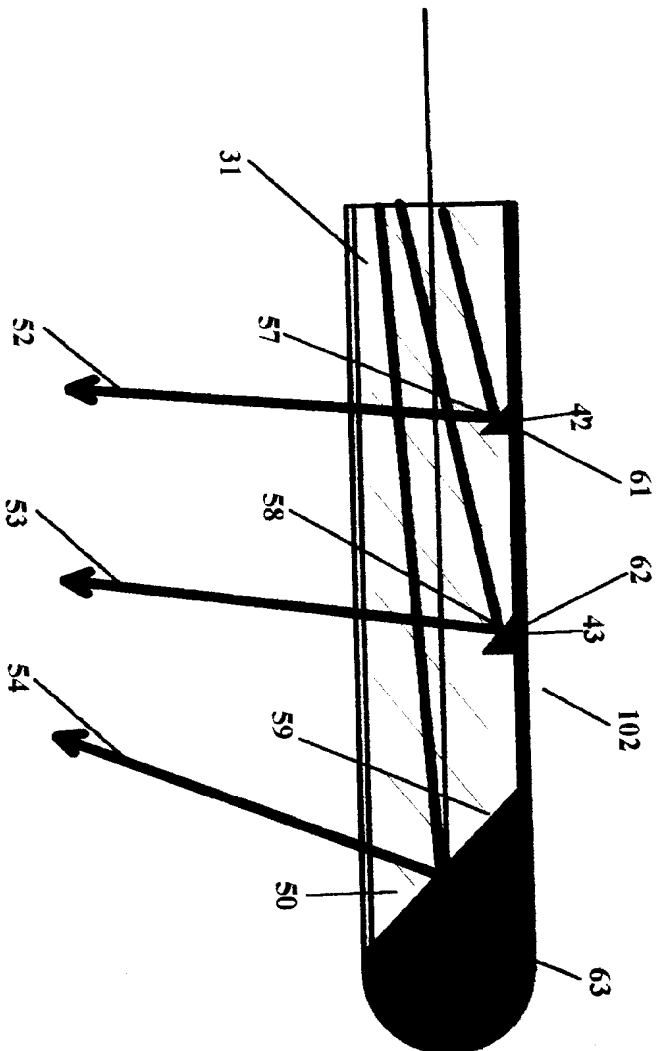


Fig 3

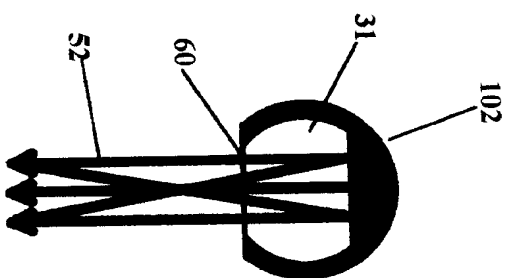


Fig 4

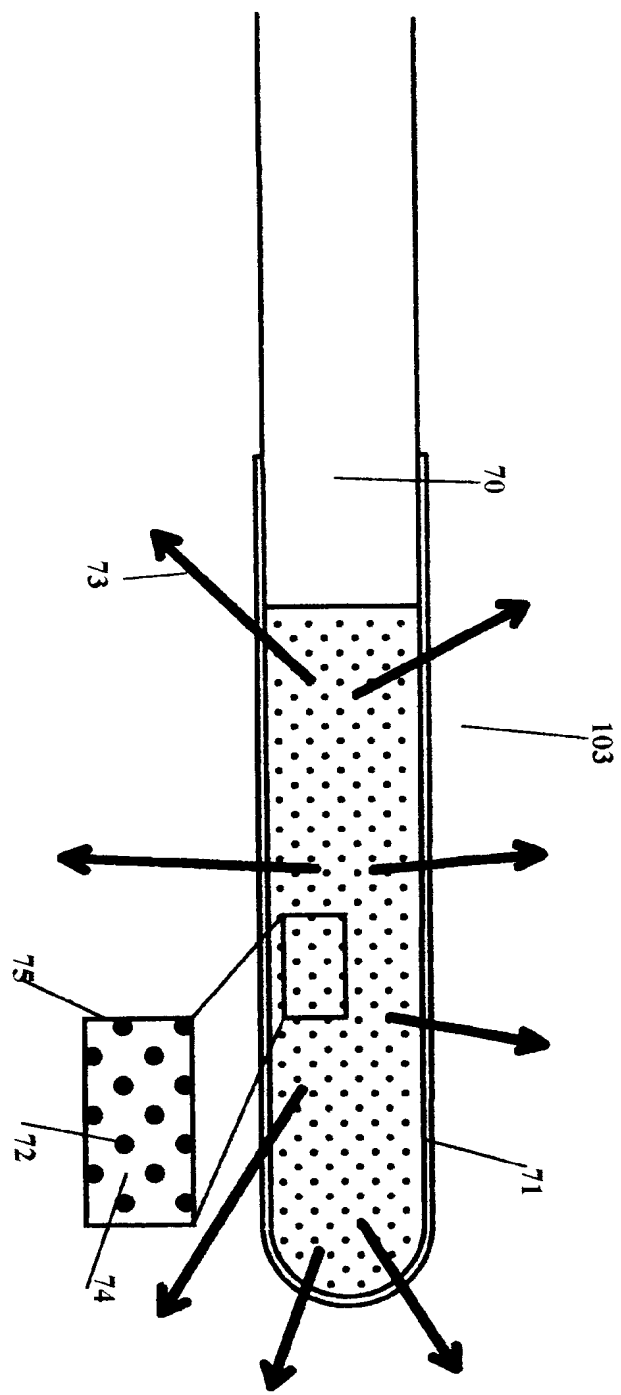


Fig 5

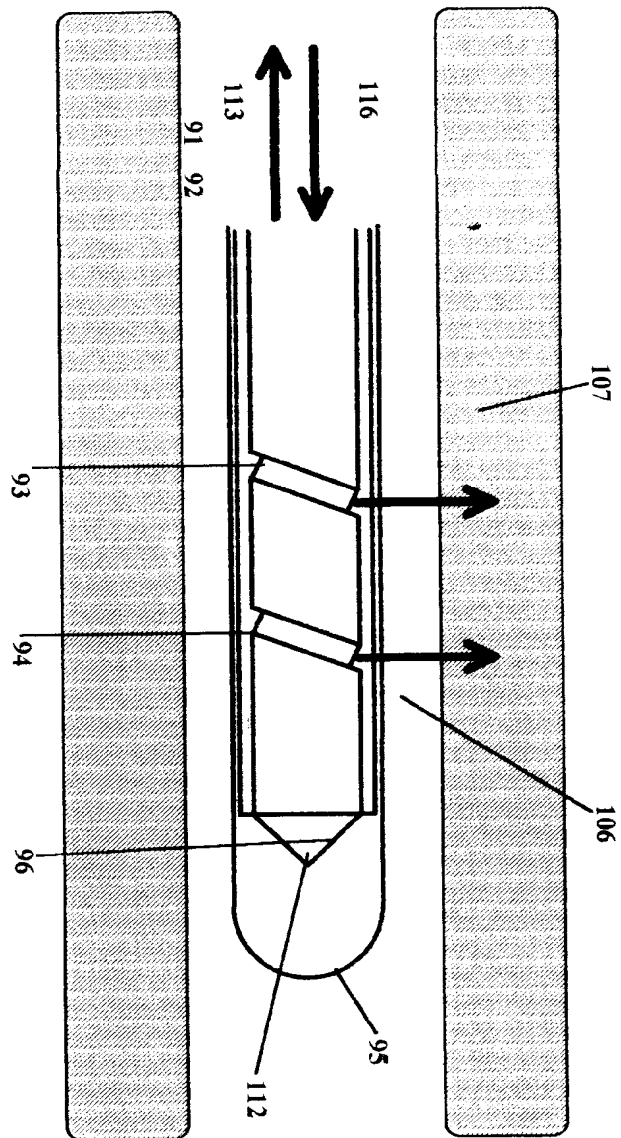


Fig 6

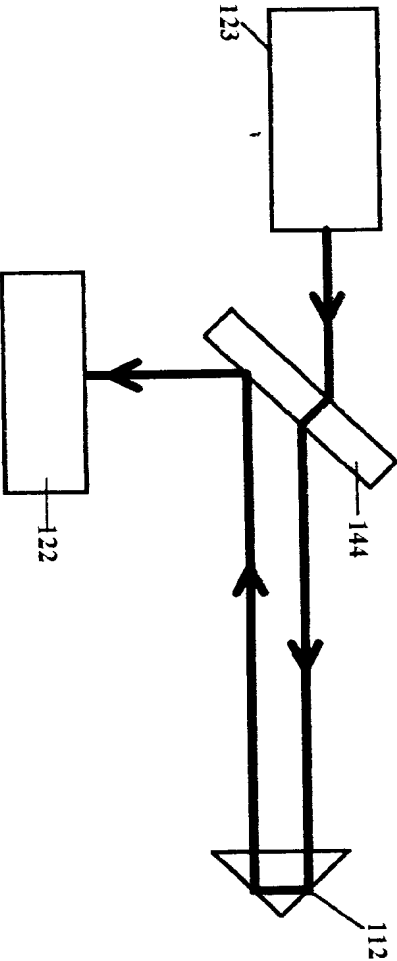


Fig 7



European Patent
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PARTIAL EUROPEAN SEARCH REPORT

Application Number

which under Rule 45 of the European Patent Convention EP 93 11 0695
shall be considered, for the purposes of subsequent
proceedings, as the European search report

DOCUMENTS CONSIDERED TO BE RELEVANT			
Category	Citation of document with indication, where appropriate, of relevant passages	Relevant to claim	CLASSIFICATION OF THE APPLICATION (Int.Cl.5)
A	DE-A-39 26 353 (K K MORITA SEISAKUSHO) * column 10, paragraph 3; figures 2C,12B,13H * ---	1	A61B17/36 G02B6/28 G02B6/36
A	WO-A-90 02349 (RAYNET) * figures 1,4,5,6 * ---	1	
A	WO-A-91 06251 (SURGILASE) * page 5, paragraph 3 * ---	1	
A	EP-A-0 292 621 (SURGICAL LASER) ---		
A	US-A-4 625 724 (SUZUKI) ---		
A	EP-A-0 182 689 (MEDICAL LASER R&D) ---		
A	EP-A-0 073 617 (PEMBERY) -----		
			TECHNICAL FIELDS SEARCHED (Int.Cl.5)
			A61B G02B
INCOMPLETE SEARCH			
<p>The Search Division considers that the present European patent application does not comply with the provisions of the European Patent Convention to such an extent that it is not possible to carry out a meaningful search into the state of the art on the basis of some of the claims</p> <p>Claims searched completely : Claims searched incompletely : Claims not searched : Reason for the limitation of the search:</p> <p>see sheet C</p>			
Place of search		Date of completion of the search	Examiner
THE HAGUE		11 February 1994	Barton, S
CATEGORY OF CITED DOCUMENTS			
X : particularly relevant if taken alone Y : particularly relevant if combined with another document of the same category A : technological background O : non-written disclosure P : intermediate document		T : theory or principle underlying the invention E : earlier patent document, but published on, or after the filing date D : document cited in the application L : document cited for other reasons & : member of the same patent family, corresponding document	



EP 93 11 0695 -C-

INCOMPLETE SEARCH

Claims searched completely : 1-12
Claims not searched : 13-20

(19)



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Remarks:

A request for correction in the last line of claim 27 has been filed pursuant to Rule 88 EPC. A decision on the request will be taken during the proceedings before the Examining Division (Guidelines for Examination in the EPO, A-V, 3.).

(54) Method and apparatus for generating bright light sources

(57) A method and apparatus for generating bright light sources includes a first plurality of semiconductor lasers in which each of the first plurality of semiconductor lasers emits a beam of light of a first frequency; a second plurality of semiconductor lasers in which each of the second plurality of semiconductor lasers emits a beam of light of a second frequency; a first bundle of optical fibers for receiving the light from ones of the first plurality of semiconductor lasers and providing a first combined laser beam of a predetermined shape and having a cross-sectional area and a solid angle; a second bundle

of optical fibers for receiving the light from ones of the second plurality of semiconductor lasers and providing a second combined laser beam of a predetermined shape and having a cross-sectional area and a solid angle; and frequency responsive optics for combining the first and second combined laser beams into a composite laser beam, the composite laser beam having substantially the same product of cross-sectional area and solid angle as each of the first and second combined beams.

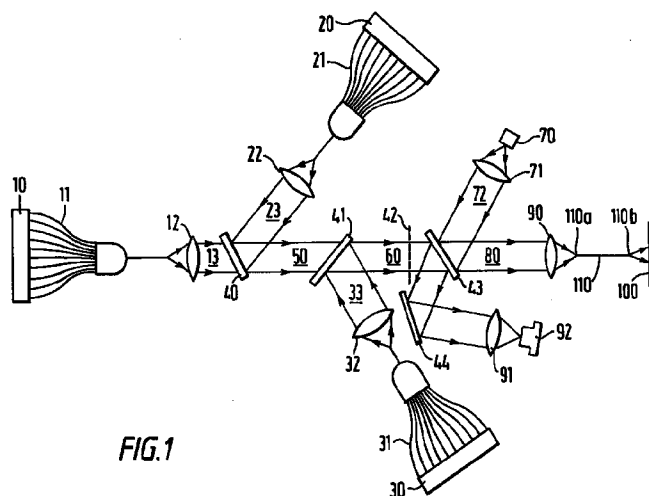


FIG.1

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Description

The present invention relates generally to a method and apparatus for generating bright light sources and has particular, although not exclusive relevance to a method and apparatus for generating bright light sources using semiconductor lasers of different wavelengths to perform medical laser applications.

In many applications, particularly medical laser applications, it is necessary to concentrate relatively high laser power onto a small area, e.g., for ablating or cutting tissue. This application of lasers requires that the laser source be of high brightness; that is, it must emit a high power beam of small area and small divergence. Brightness is directly proportional to the output power and inversely proportional to the source area and also to the solid angle of the light beam. (Brightness = power / (Source Area x Solid Angle)) It is therefore difficult to increase the brightness of a source by increasing the output power, since this also tends to increase the source area and/or the solid angle of the light beam such that the net effect on the brightness is relatively small. For example, suppose the object is to double the brightness of a particular laser that emits 1 watt of power into a certain solid angle. By simply taking another such laser and placing it adjacent the first, the result is two watts of power into essentially the same solid angle, but the total cross-sectional area of the new source is also twice as large such that the brightness of the combined source is essentially equal to the brightness of each separate laser (not double the brightness).

It has been proposed to use a plurality of laser diodes to provide a higher power laser diode light source from an optical fiber. An example of such a laser diode light source is disclosed in UK Patent Application GB 2256503A.

The prior art teaches to increase the brightness of a light source by using a polarized beam combiner on two beams of light of orthogonal polarizations from two laser diodes, thus combining the two beams into a single composite beam. The prior art further teaches to feed this composite beam into a fiber optic cable and then to bundle a group of these fiber optic cables to form a larger more powerful beam of light. This method suffers from a number of problems some of which include: the power output being limited by being able to combine only two laser diodes with each polarized beam combiner, and the multitude of polarized lasers and optics being very complex and expensive (this method requires a large number of beam combiner pairs to achieve a desired high power).

Thus there exists a need for a more efficient way to combine a plurality of laser diodes to form a relatively high brightness laser light source for performing medical laser applications.

It is thus an object of the present invention to provide a method of and an apparatus for generating a beam of bright light for medical laser applications.

It is another object of the present invention to provide a method of and an apparatus for generating a beam of bright light by efficiently combining a plurality of semiconductor lasers.

It is another object of the present invention to provide a method of and an apparatus for generating a beam of bright light by combining a plurality of semiconductor lasers of different frequencies.

It is yet another object of the present invention to provide a method of performing medical laser applications by generating a first laser beam of a first frequency, generating a second laser beam of a second frequency, combining the first and second laser beams into a composite laser beam then shining the composite laser beam into tissue.

Another object of the present invention is to provide an apparatus for generating a beam of bright light utilizing semiconductor lasers of different frequencies, bundles of optical fibers to form composite single frequency laser beams from the lasers of different frequencies respectively and frequency responsive optics for combining the composite laser beams of the different frequencies into a combined laser beam.

The foregoing objects are attained by the invention, which provides a method of and an apparatus for generating a bright light source including a first and a second plurality of semiconductor lasers in which each of the first and second plurality of semiconductor lasers respectively emits a beam of a first and second frequency. A first and a second bundle of optical fibers may respectively be used for receiving the light from ones of the first and second plurality of semiconductor lasers and for providing a first and a second combined laser beam of a predetermined shape and having a cross-sectional area and a solid angle. Frequency responsive optics are used for combining the first and second combined laser beams into a composite laser beam having substantially the same product of cross-sectional area and solid angle as each of the first and second combined beams.

The invention will next be described in connection with certain illustrated embodiments; however, it should be clear to those skilled in the art that various modifications, additions and subtractions can be made without departing from the spirit or scope of the claims.

For a fuller understanding of the nature and objects of the invention, an embodiment of the invention will now be described by way of example only with reference to the accompanying drawings in which:

FIG 1. illustrates one form of apparatus constructed in accordance with an embodiment of the present invention.

FIG. 1 depicts a bright light source in accordance with an embodiment of the invention such that the laser beam from the source is applied onto a working surface, such as tissue to be subjected to a medical laser treatment. The input power of each light source, as defined by a plurality of laser diodes, is of the order of many watts or tens of watts. The

produced composite beam is transmitted a relatively short distance (e.g., up to a few meters) such that the losses are very low or negligible, and therefore, the power of the composite beam is substantially equal to the sum of the powers of the input beams, again many watts or tens of watts.

The above technique for producing a bright light source is applicable generally with respect to a wide variety of light sources, but is particularly attractive in the case of laser diodes. It is known that laser diodes are small, efficient, easy to drive and modulate, very reliable, and of relatively low cost. However, the brightness of the presently known laser diodes, including the high power laser diodes, is generally far below the brightness of gas lasers or solid state lasers. Thus, the resulting power per unit area, per unit solid angle of laser light radiation generated by a laser diode, is insufficient for many medical laser applications.

The present invention exploits the above advantageous characteristics of laser diodes, and the further characteristic that laser diodes can be made to lase in a wide range of wavelengths. For example, the presently known high power laser diodes made of GaAlAs emit light in a range of 780 to 880 nm, and diodes made of InGaAs emit light in the range of 920 to 980 nm. The above-described techniques for increasing the brightness of a beam of light thus enables laser diodes to be used in applications, such as in medical laser applications, requiring high brightness outputs.

Thus, in the preferred embodiment of the invention, the source beam generators include a plurality of laser diodes of different wavelengths. In the described example, these generators include a first plurality of diodes 10 that produce laser beams of 780 nm, a second plurality of laser diodes 20 that produce laser beams of 860 nm, and a third plurality of laser diodes 30 that produce laser beams of 960 nm. It will be obvious to one skilled in the art that the generators could also be single high powered laser diodes of the previously mentioned frequencies. The power of each laser beam formed by combining the first plurality of laser diodes, the second plurality of laser diodes and the third plurality of laser diodes respectively is at least 10 watts, preferably about 15 watts. Thus the power of the composite output beam is approximately 45 watts, sufficient for use in many surgical laser applications. It will be appreciated that if higher powers are desired, more than three source beams can be combined in accordance with the present invention.

According to further features of the preferred embodiment, the plurality of beams are combined into the single composite beam such that the product of the cross-sectional area and the solid angle, of the single composite beam is substantially the same as the product in each of the source beams.

The apparatus illustrated in the drawing includes a first plurality of laser diodes 10, a second plurality of laser diodes 20 and a third plurality of laser diodes 30 generating laser beams of different wavelengths, e.g., of 780, 860 and 960 nm, respectively. As the laser energy from each plurality of laser diodes is emitted from a relatively wide area (typically about 10mm), the output from each plurality of laser diodes is directed to a fibre bundle 11, 21 and 31 respectively, which is tightly packed at the output end, e.g., to a circle of about 1 mm in diameter. Lenses 12, 22 and 32, respectively, collimate the light from each fibre bundle to produce three essentially collimated beams 13, 23 and 33, respectively.

The illustrated apparatus further includes frequency responsive optics in the form of a plurality of wavelength combiners 40, 41 for combining the plurality of light beams 13, 23 and 33, respectively, into a single composite beam. Thus, a first combiner 40 combines beams 13 and 23 into a composite beam 50; and a second combiner 41 combines beams 50 and 33 into a single composite beam 60.

In the illustrated example, combiners 40 and 41 are both dichroic elements, e.g., interference filters, which are highly transparent to one wavelength band and highly reflective to the other wavelength band. Thus, dichroic combiner 40 is highly transmissive with respect to the wavelength of laser beam 13 and is highly reflective with respect to the wavelength of laser beam 23; whereas combiner 41 is highly transmissive with respect to the wavelengths of both laser beams 13 and 23, and is highly reflective with respect to the wavelength of laser beam 33.

Each of the plurality of laser diodes 10, 20 and 30, should output at least one watt of power, preferably at least 10 watts of power in medical laser applications of the invention. In the illustrated example, each outputs 15 watts of power, so that the power of the single composite beam 60 is approximately 45 watts. This is normally sufficient to permit the laser beam to be used for many medical laser applications.

The illustrated apparatus further includes a safety shutter 42 in the path of the single composite beam 60, in order to selectively block the transmission of the single composite beam to the working surface 100.

The illustrated system further includes a fourth laser 70 which generates a laser beam of visible light and of low power, to serve as a visible aiming beam to be projected onto the working surface 100 in order to permit aiming the high powered composite beam 60. In the illustrated example, the aiming beam laser 70 is a laser diode generating a laser beam of 635 nm wavelength and of a few milliwatts of power. This laser beam is collimated by a lens 71 to produce a collimated beam 72 which is combined by a third dichroic combiner 43 with the high power composite beam 60 from the three pluralities of laser diodes 10, 20 and 30. Dichroic combiner 43 would thus be highly transmissive with respect to the wavelengths of the three pluralities of laser diodes 10, 20 and 30 but highly reflective with respect to the wavelength of the aiming laser 70. The output of the dichroic combiner 43 is thus a single composite beam 80 including the outputs of the three pluralities of laser diodes 10, 20 and 30 and the output of the aiming beam laser 70.

The output 80 of dichroic combiner 43 is then directed, via a focusing lens 90, into the inlet end 110a of a guiding optical fibre 110. The single composite beam exits from the outlet end 110b of the optical fibre 110 onto the working surface 100.

In the described apparatus, the combiners 40, 41 and 43, as well as the optics of the apparatus, are such that the product of the cross-sectional area and the solid angle of the single composite beam 80 outputted by combiner 43 is substantially the same as in each of the beams exiting from the plurality of laser diodes 10, 20 and 30 via their respective bundles.

Thus, the outlet end 110b of the guiding optical fibre 110 becomes a light source of high brightness, almost three times that of each of the pluralities of laser diodes, thereby enabling the use of laser sources, particularly laser diodes, for applications requiring high power and high brightness, such as medical laser applications.

While the dichroic combiner 43 is highly transmissive of the wavelengths of the three essentially collimated beams 13, 23 and 33 in the composite beam 60, a small fraction of the energy in the composite beam is reflected by the combiner 43. This small fraction reflected by combiner 43 is used for monitoring the power produced in the composite beam 80. For this purpose, the small fraction of the laser power reflected by combiner 43 is reflected by a mirror 44 through a focusing lens 91 onto a power detector 92 for measuring the laser energy so reflected from the combiner 43. Since the proportion of laser energy reflected by combiner 43 to the laser energy passing through the combiner 43 is known, depending on the characteristics of the combiner, the output of power detector 92 also provides a measurement of the total energy in the composite beam 80 applied, via optical fibre 110, onto the working surface 100.

It will be appreciated that whereas the illustrated system includes dichroic combiners, other types of wavelength combiners may be used for multiplexing the laser beams, e.g., a holographic element or grating, as known in the prior art. It will also be appreciated that whereas the illustrated apparatus shows combining three light sources, namely plurality of laser diodes 10, 20 and 30 together with the aiming beam laser 70, only two, or more than three, light sources could be combined, according to the requirements of a particular application.

It will thus be seen that the invention efficiently attains the objects set forth above, among those made apparent from the preceding description. In particular, the invention provides a method of and an apparatus for generating bright light sources using semiconductor lasers of different frequencies to perform medical laser applications. Those skilled in the art will appreciate that the configuration depicted in Fig. 1 efficiently and effectively attains a bright light source having a high power emitted from a tight radius into a confined solid angle.

It will be understood that changes may be made in the above construction and in the foregoing sequences of operation without departing from the scope of the invention. For instance it will be appreciated that the invention could be advantageously used in other applications requiring bright light sources, e.g., metal welding. It is accordingly intended that all matter contained in the above description or shown in the accompanying drawings be interpreted as illustrative rather than in a limiting sense.

It is also to be understood that the following claims are intended to cover all of the generic and specific features of the invention as described herein, and all statements of the scope of the invention which, as a matter of language, might be said to fall therebetween.

Claims

1. A method of producing a light source including the steps of:
 - generating a first laser beam of a first frequency;
 - generating a second laser beam of a second frequency and
 - combining said first and second laser beams into a composite laser beam.
2. A method according to claim 1 for performing medical laser applications including the step of:
 - shining said composite laser beam into tissue to perform medical laser applications.
3. The method according to claim 1 or 2, further including the steps of:
 - generating a third laser beam of a third frequency;
 - combining said third laser beam with said first and second laser beams into said composite laser beam.
4. The method according to anyone of claims 1 to 3, wherein
 - at least said first and second laser beams are generated by diode lasers.
5. The method according to any one of claims 1 to 4, wherein
 - at least said first laser beam, said second laser beam and said composite laser beam each has a cross-sectional area and a solid angle such that the product of the cross-sectional area and the solid angle of the composite laser beam is substantially the same as that of each of said first and second laser beams.
6. The method according to any one of claims 1 to 5, wherein
 - at least said first and second laser beams are combined into said composite beam by a dichroic combiner.

7. The method according to any one of claims 1 to 5, wherein
at least said first and second laser beams are combined into said composite beam by a holographic element.
8. The method according to any one of claims 1 to 5, wherein
at least said first and second laser beams are combined into said composite beam by a grating.
9. The method according to claim 4, wherein
said diode lasers are at least one watt of power each.
10. The method according to any one of claims 1 to 9, wherein
a small fraction of the composite beam is split-off and is directed to a power detector for measuring the power of said composite beam.
11. The method according to claim 3, wherein
said first laser beam is of a wavelength of 780 nm and of a power of at least 10 watts;
said second laser beam is of a wavelength of 860 nm and of a power of at least 10 watts; and
said third laser beam is of a wavelength of 960 nm and of a power of at least 10 watts.
12. The method according to any one of claims 1 to 11, wherein
an aiming beam of visible light and of a different wavelength than said first, second and third laser beams is also generated and combined into said composite beam.
13. The method according to claim 12, wherein
said aiming beam is of a wavelength of 635 nm and of a power of a few milliwatts.
14. The method according to any one of claims 1 to 13, wherein said composite beam is directed into said tissue by an optical fibre.
15. A bright light source including:
a first plurality of semiconductor lasers in which each of said first plurality of semiconductor lasers emits a beam of light of a first frequency;
a second plurality of semiconductor lasers in which each of said second plurality of semiconductor lasers emits a beam of light of a second frequency;
a first bundle of optical fibers for receiving the light from ones of said first plurality of semiconductor lasers and providing a first combined laser beam of a predetermined shape and having a cross-sectional area and a solid angle;
a second bundle of optical fibers for receiving the light from ones of said second plurality of semiconductor lasers and providing a second combined laser beam of a predetermined shape and having a cross-sectional area and a solid angle; and
frequency responsive optics for combining said first and second combined laser beams into a composite laser beam, said composite laser beam having substantially the same product of cross-sectional area and solid angle as each of said first and second combined beams.
16. A bright light source as defined in claim 15, wherein
said first plurality of semiconductor lasers are arranged in a first monolithic array; and
said second plurality of semiconductor lasers are arranged in a second monolithic array.
17. A bright light source as defined in claim 15 or 16 further including:
a third plurality of semiconductor lasers in which each of said third plurality of semiconductor lasers emits a beam of light of a third frequency;
a third bundle of optical fibers for receiving the light from ones of said third plurality of semiconductor lasers and providing a third combined laser beam of a predetermined shape and having a cross-sectional area and a solid angle.
18. The apparatus according to any one of claims 15 to 17, wherein
said semiconductor lasers are high power diode lasers.
19. The apparatus according to any one of claims 15 to 18, wherein
said frequency responsive optics include a dichroic combiner.

20. The apparatus according to any one of claims 15 to 18, wherein
said frequency responsive optics includes a grating.

21. The apparatus according to any one of claims 15 to 18, wherein
said frequency responsive optics include a holographic element.

22. The apparatus according to any one of claims 15 to 21, further including
a power detector for receiving a small fraction of the composite beam and for measuring the power of the
combined beam.

23. The apparatus according to any one of claims 15 to 22, wherein
each of said combined beams is at least 10 watts of power.

24. The apparatus according to any one of claims 15 to 23, wherein
said first plurality of semiconductor lasers each generates a laser beam of between 780 nm and 880 nm
inclusive; and
said second plurality of semiconductor lasers each generates a laser beam of between 920 nm and 940 nm
inclusive.

25. The apparatus according to any one of claims 15 to 24, including
a light source for generating an aiming beam of visible light and of a different wavelength than said plurality
of semiconductor lasers; and
a multiplexer for multiplexing said aiming beam into said combined laser beam.

26. The apparatus according to claim 25, wherein said further light source generates an aiming beam of 635 nm and
of a few milliwatts of power.

27. A light source comprising at least two semiconductor lasers effective to generate first and second laser beams of
different frequencies; and means for combining said second laser beams into a composite laser beam.

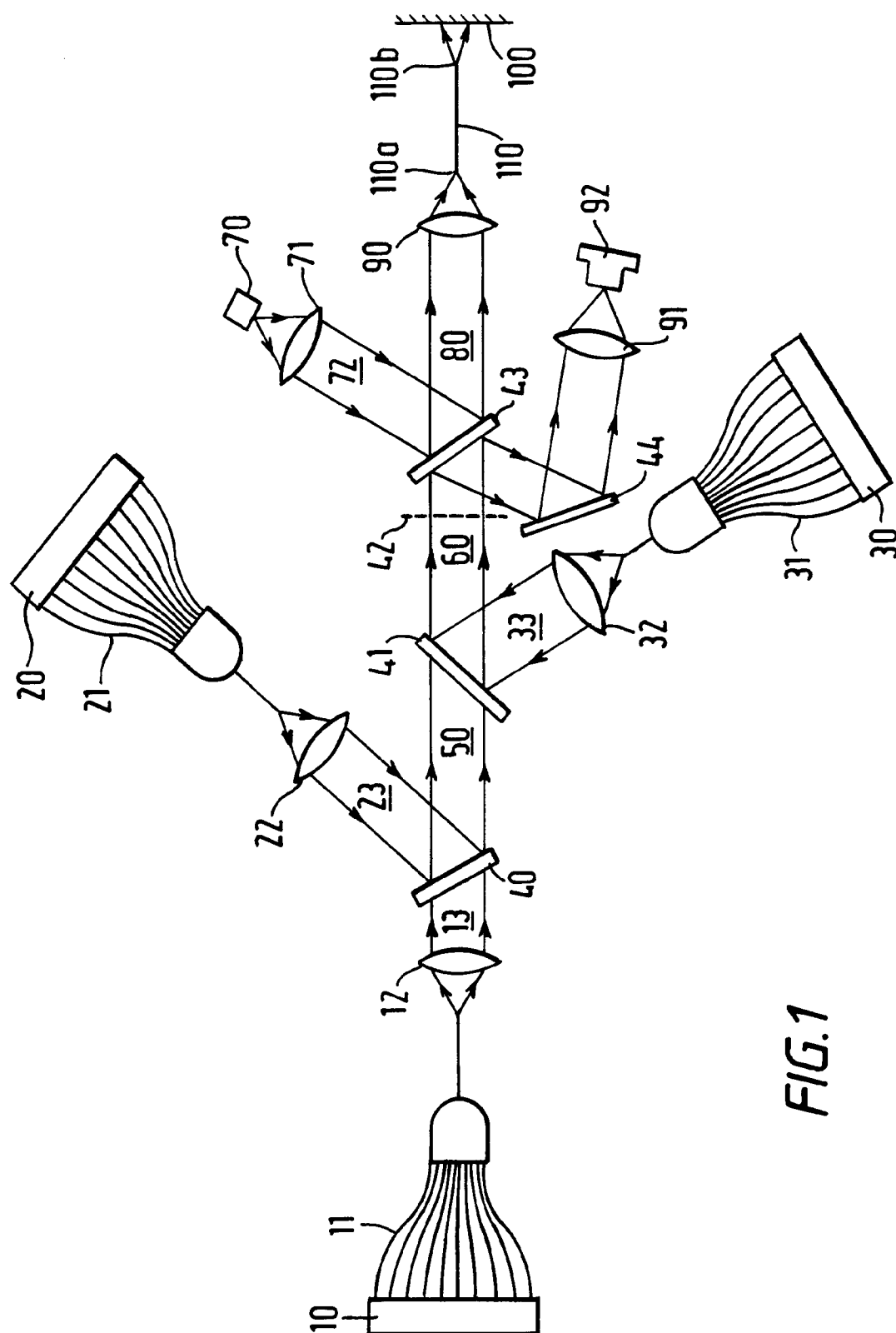


FIG. 1



European Patent
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EUROPEAN SEARCH REPORT

Application Number
EP 94 30 7879

DOCUMENTS CONSIDERED TO BE RELEVANT			
Category	Citation of document with indication, where appropriate, of relevant passages	Relevant to claim	CLASSIFICATION OF THE APPLICATION (Int.Cl.6)
X	WO-A-93 21843 (COHERENT INC) * page 17, line 5 - page 18, line 17 * ---	1-6,9	H01S3/25 A61B17/36 G02B6/293
X	US-A-4 847 479 (G.L.CLARK) * figure 1 * ---	1,4,7-10	
Y	WO-A-91 12641 (SCIENTIFIC GENERICS LTD) * figure 3 * ---	15-21	
D	& GB-A-2 256 503 ---		
X	PATENT ABSTRACTS OF JAPAN vol. 6, no. 144 (E-122) (1022) 3 August 1982 & JP-A-57 068 098 (MITSUBISHI DENKI KK) 26 April 1982 * abstract * ---	1,4,14,27	
Y	* figures 1,2 * ---	15-21	
A	WO-A-91 01056 (AUSTRALIAN ELECTRO OPTICS PTY LTD) -----	15	
			TECHNICAL FIELDS SEARCHED (Int.Cl.6)
			H01S A61B G02B
The present search report has been drawn up for all claims			
Place of search THE HAGUE		Date of completion of the search 18 April 1995	Examiner Galanti, M
CATEGORY OF CITED DOCUMENTS		T : theory or principle underlying the invention E : earlier patent document, but published on, or after the filing date D : document cited in the application L : document cited for other reasons & : member of the same patent family, corresponding document	
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(54) Method and apparatus for therapeutic electromagnetic treatment

(57) A therapeutic treatment device for treating a treatment region comprising an incoherent, pulsed, light source operable to provide a light output for treatment, a power supply connected to the light source, a housing including a reflector and having an opening, wherein the light source is disposed within the housing and the reflector reflects light from the light source to the opening. A flexible light guide is disposed between the opening and the treatment region, wherein the light guide receives the incoherent light from the light source and transmits the light to the treatment region and the light source, reflector and light guide cooperate to provide between 6 and 100 J/cm² to the skin.

The light guide transmits light having a predetermined angular divergence, wherein the divergence is selected in response to a desired treatment depth.

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Description

The present invention relates generally to the art of therapeutic electromagnetic treatment and more specifically to a method and apparatus for utilizing a spatially extended pulsed light source such as a flashlamp (flash tube) for such a treatment or, efficiently focusing light from the flashlamp into optical fibers for therapeutic treatment or other applications.

This application is a continuation-in-part of prior co-pending United States application Serial No. 07/964,210, filed October 20, 1992, entitled "Method And Apparatus For Therapeutic Electromagnetic Treatment."

It is known in the prior art to use electromagnetic radiation in medical application for therapeutic uses such as treatment of skin disorders. For example, U.S. Patent No. 4,298,005 to Mutzhas describes a continuous ultraviolet lamp with cosmetic, photobiological, and photochemical applications. A treatment based on using the UV portion of the spectrum and its photochemical interaction with the skin is described. The power delivered to the skin using Mutzhas' lamp is described as 150W/m^2 , which does not have a significant effect on skin temperature.

In addition to prior art treatment involving UV light, lasers have been used for dermatological procedures, including Argon lasers, CO_2 lasers, Nd(Yag) lasers, Copper vapor lasers, ruby lasers and dye lasers. For example, U.S. Patent No. 4,829,262 to Furumoto, describes a method of constructing a dye laser used in dermatology applications. Two skin conditions which may be treated by laser radiation are external skin irregularities such as local differences in the pigmentation or structure of the skin, and vascular disorders lying deeper under the skin which cause a variety of skin abnormalities including port wine stains, telangiectasias, leg veins and cherry and spider angiomas. Laser treatment of these skin disorders generally includes localized heating of the treatment area by absorption of laser radiation. Heating the skin changes or corrects the skin disorder and causes the full or partial disappearance of the skin abnormality.

Certain external disorders such as pigmented lesions can also be treated by heating the skin very fast to a high enough temperature to evaporate parts of the skin. Deeper-lying vascular disorders are more typically treated by heating the blood to a high enough temperature to cause it to coagulate. The disorder will then eventually disappear. To control the treatment depth a pulsed radiation source is often used. The depth the heat penetrates in the blood vessel is controlled by controlling the pulse width of the radiation source. The absorption and scattering coefficients of the skin also affect the heat penetration. These coefficients are a function of the constituents of skin and the wavelength of the radiation. Specifically, the absorption coefficient of light in the epidermis and dermis tends to be a slowly varying, monotonically decreasing function of wavelength. Thus, the wavelength of the light should be chosen so that the absorption coefficient is optimized for the particular skin condition and vessel size being treated.

The effectiveness of lasers for applications such as tattoo removal and removal of birth and age marks is diminished because lasers are monochromatic. A laser of a given wavelength may be effectively used to treat a first type of skin pigmentation disorder, but, if the specific wavelength of the laser is not absorbed efficiently by skin having a second type of disorder, it will be ineffective for the second type of skin disorder. Also, lasers are usually complicated, expensive to manufacture, large for the amount of power delivered, unreliable and difficult to maintain.

The wavelength of the light also affects vascular disorder treatment because blood content in the vicinity of the vascular disorders varies, and blood content affects the absorption coefficient of the treatment area. Oxyhemoglobin is the main chromophore which controls the optical properties of blood and has strong absorption bands in the visible region. More particularly, the strongest absorption peak of oxyhemoglobin occurs at 418nm and has a band-width of 60nm. Two additional absorption peaks with lower absorption coefficients occur at 542 and 577nm. The total band-width of these two peaks is on the order of 100nm. Additionally, light in the wavelength range of 500 to 600nm is desirable for the treatment of blood vessel disorders of the skin since it is absorbed by the blood and penetrates through the skin. Longer wavelengths up to 1000nm are also effective since they can penetrate deeper into the skin, heat the surrounding tissue and, if the pulse-width is long enough, contribute to heating the blood vessel by thermal conductivity. Also, longer wavelengths are effective for treatment of larger diameter vessels because the lower absorption coefficient is compensated for by the longer path of light in the vessel.

Accordingly, a wide band electromagnetic radiation source that covers the near UV and the visible portion of the spectrum would be desirable for treatment of external skin and vascular disorders. The overall range of wavelengths of the light source should be sufficient to optimize treatment for any of a number of applications. Such a therapeutic electromagnetic radiation device should also be capable of providing an optimal wavelength range within the overall range for the specific disorder being treated. The intensity of the light should be sufficient to cause the required thermal effect by raising the temperature of the treatment area to the required temperature. Also, the pulse-width should be variable over a wide enough range so as to achieve the optimal penetration depth for each application. Therefore, it is desirable to provide a light source having a wide range of wavelengths, which can be selected according to the required skin treatment, with a controlled pulse-width and a high enough energy density for application to the affected area.

Pulsed non-laser type light sources such as linear flashlamps provide these benefits. The intensity of the emitted light can be made high enough to achieve the required thermal effects. The pulse-width can be varied over a wide range so that control of thermal depth penetration can be accomplished. The typical spectrum covers the visible and ultraviolet range and the optical bands most effective for specific applications can be selected, or enhanced using fluorescent

materials. Moreover, non-laser type light sources such as flashlamps are much simpler and easier to manufacture than lasers, are significantly less expensive for the same output power and have the potential of being more efficient and more reliable. They have a wide spectral range that can be optimized for a variety of specific skin treatment applications. These sources also have a pulse length that can be varied over a wide range which is critical for the different types of skin treatments.

In addition to being used for treating skin disorders, lasers have been used for invasive medical procedures such as lithotripsy and removal of blood vessel blockage. In such invasive procedures laser light is coupled to optical fibers and delivered through the fiber to the treatment area. In lithotripsy the fiber delivers light from a pulsed laser to a kidney or gallstone and the light interaction with the stone creates a shock wave which pulverizes the stone. To remove blood vessel blockage the light is coupled to the blockage by the fiber and disintegrates the blockage. In either case the shortcomings of lasers discussed above with respect to laser skin treatment are present. Accordingly, a treatment device for lithotripsy and blockage removal utilizing a flashlamp would be desirable.

To effectively treat an area the light from the source must be focussed on the treatment area. Coupling pulsed laser light into optical fibers in medicine is quite common. The prior art describes coupling isotropic incoherent point sources such as CW lamps into small optical fibers. For example, U. S. Patent No. 4,757,431, issued July 12, 1988, to Cross, et al. discloses a method for focusing incoherent point sources with small filaments or an arc lamp with an electrode separation of 2mm into a small area. Point (or small) sources are relatively easy to focus without large losses in energy because of the small size of the source. Also, U. S. Patent No. 4,022,534, issued May 10, 1977, to Kishner discloses light produced by a flash tube and the collection of only a small portion of the light emitted by the tube into an optical fiber.

However, the large dimension of an extended source such as a flashlamp makes it difficult to focus large fractions of its energy into small areas. Coupling into optical fibers is even more difficult since not only must a high energy density be achieved, but the angular distribution of the light has to be such that trapping in the optical fiber can be accomplished. Thus, it is desirable to have a system for coupling the output of a high intensity, extended, pulsed light source into an optical fiber.

According to a first embodiment of the invention a therapeutic treatment device comprises a housing and an incoherent light source, suitably a flashlamp, operable to provide a pulsed light output for treatment, disposed in the housing. The housing has an opening and is suitable for being disposed adjacent a skin treatment area. A reflector is mounted within the housing proximate the light source, and at least one optical filter is mounted proximate the opening in the housing. An iris is mounted coextensively with the opening. Power to the lamp is provided by a variable pulse width pulse forming circuit. Thus, the treatment device provides controlled density, filtered, pulsed light output through an opening in the housing to a skin area for treatment.

According to a second embodiment of the invention a method of treatment with light energy comprises the steps of providing a high power, pulsed light output from a non-laser, incoherent light source and directing the pulsed light output to a treatment area. The pulse width of the light output is controlled and focussed so that the power density of the light is controlled. Also, the light is filtered to control the spectrum of the light.

According to a third embodiment of the invention a coupler comprises an incoherent light source such as a toroidal flashlamp. A reflector is disposed around the incoherent light source and at least one optical fiber or light guide. The fiber has an end disposed within the reflector. This end collects the light from the circular lamp. In a similar coupling configuration fibers may be provided, along with a linear to circular fiber transfer unit disposed to receive light from the light source and provide light to the optical fibers. The reflector has an elliptical cross-section in a plane parallel to the axis of the linear flash tube, and the linear flash tube is located at one focus of the ellipse while the linear to circular transfer unit is located at the other focus of the ellipse.

For a better understanding of the invention, reference is made to the accompanying drawings, in which like numerals designate corresponding elements or sections throughout, and in which:

Figure 1 is a cross-sectional view of an incoherent, pulsed light source skin treatment device;

Figure 2 is a side view of the light source of Figure 1;

Figure 3 is a schematic diagram of a pulse forming network with a variable pulse width for use with the skin treatment device of Figures 1 and 2;

Figure 4 is a cross-sectional view of a coupler for coupling light from a toroidal flash tube into an optical fiber with a conical edge;

Figure 5 is a side view of a toroidal flash tube;

Figure 6 is a top view of a toroidal flash tube;

Figure 7 shows the geometry for coupling into a conical section;

Figure 8 is a cross-sectional view of a coupler for coupling light from a toroidal flash tube into an optical fiber with a flat edge;

Figure 9 is a front sectional view of a coupler for coupling light from a linear flash tube into a circular fiber bundle;

Figure 10 is a side sectional view of the coupler of Figure 9;

Figure 11 is a front view of a coupler for coupling light from a linear flash tube into an optical fiber;
 Figure 12 is a front view of a coupler for coupling light from a linear flash tube into a doped optical fiber;
 Figure 13 is a schematic configuration of a gel skin interface with a transparent plate;
 Figure 14 shows an angular distribution of photons penetrating without using a gel;
 Figure 15 shows a light guide providing a large angular divergence;
 Figure 16 shows a light guide providing a narrow angular divergence;
 Figure 17 shows a spectra produced with a flashlamp current of 200 amps;
 Figure 18 shows a spectra produced with a flashlamp current of 200 amps; and
 Figure 19 shows a GTO driver circuit for a flashlamp.

In the various figures, like reference numerals are used to describe like components.

Before explaining at least one embodiment of the invention in detail it is to be understood that the invention is not limited in its application to the details of construction and the arrangement of the components set forth in the following description or illustrated in the drawings. The invention is capable of other embodiments or of being practiced or carried out in various ways. Also, it is to be understood that the phraseology and terminology employed herein is for the purpose of description and should not be regarded as limiting.

Referring now to Figures 1 and 2, cross-sectional and side views of an incoherent, pulsed light source skin treatment device 10 constructed and operated in accordance with the principles of the present invention are shown. The device 10 may be seen to include a housing 12, having an opening therein, a handle 13 (Figure 2 only), a light source 14 having an outer glass tube 15, an elliptical reflector 16, a set of optical filters 18, an iris 20 and a detector 22 (Figure 1 only).

Light source 14, which is mounted in housing 12, may be a typical incoherent light source such as a gas filled linear flashlamp Model No. L5568 available from ILC. The spectrum of light emitted by gas filled linear flashlamp 14 depends on current density, type of glass envelope material and gas mixture used in the tube. For large current densities (e.g., 3000 A/Cm² or more) the spectrum is similar to a black body radiation spectrum. Typically, most of the energy is emitted in the 300 to 1000nm wavelength range.

To treat a skin (or visible) disorder a required light density on the skin must be delivered. This light density can be achieved with the focusing arrangement shown in Figures 1 and 2. Figure 1 shows a cross-section view of reflector 16, also mounted in housing 12. As shown in Figure 1, the cross-section of reflector 16 in a plane is perpendicular to the axis of flashlamp 14 is an ellipse. Linear flashlamp 14 is located at one focus of the ellipse and reflector 16 is positioned in such a way that the treatment area of skin 21 is located at the other focus. The arrangement shown is similar to focusing arrangements used with lasers and efficiently couples light from flashlamp 14 to the skin. This arrangement should not, however, be considered limiting. Elliptical reflector 16 may be a metallic reflector, typically polished aluminum which is an easily machinable reflector and has a very high reflectivity in the visible, and the UV range of the spectrum can be used. Other bare or coated metals can also be used for this purpose.

Optical and neutral density filters 18 are mounted in housing 12 near the treatment area and may be moved into the beam or out of the beam to control the spectrum and intensity of the light. Typically, 50 to 100nm bandwidth filters, as well as low cutoff filters in the visible and ultraviolet portions of the spectrum, are used. In some procedures it is desirable to use most of the spectrum, with only the UV portion being cut off. In other applications, mainly for deeper penetration, it is preferable to use narrower bandwidths. The bandwidth filters and the cutoff filters are readily available commercially.

Glass tube 15 is located coaxially with flashlamp 14 and has fluorescent material deposited on it. Glass tube 15 will typically be used for treatment of coagulation of blood vessels to optimize the energy efficiency of device 10. The fluorescent material can be chosen to absorb the UV portion of the spectrum of flashlamp 14 and generate light in the 500 to 650nm range that is optimized for absorption in the blood. Similar materials are coated on the inner walls of commercial fluorescent lamps. A typical material used to generate "warm" white light in fluorescent lamps has a conversion efficiency of 80%, has a peak emission wavelength of 570nm and has a bandwidth of 70nm and is useful for absorption in blood. The few millisecond decay time of these phosphors is consistent with long pulses that are required for the treatment of blood vessels.

Other shapes or configurations of flashlamp 14 such as circular, helical, short arc and multiple linear flashlamps may be used. Reflector 16 may have other designs such as parabolic or circular reflectors. The light source can also be used without a reflector and the required energy and power density may be achieved by locating light source 14 in close proximity to the treatment area.

Iris 20 is mounted in housing 12 between optical filters 18 and the treatment area and controls the length and the width of the exposed area, i.e. by collimating the output of flashlamp 14. The length of flashlamp 14 controls the maximum length that can be exposed. Typically an 8cm long (arc length) tube will be used and only the central 5cm of the tube is exposed. Using the central 5cm assures a high degree of uniformity of energy density in the exposed skin area. Thus, in this embodiment the iris 20 (also called a collimator) will enable exposure of skin areas of a maximum length of 5cm. The iris 20 may be closed to provide a minimum exposure length of one millimeter. Similarly, the width of the

exposed skin area can be controlled in the range of 1 to 5mm for a 5mm wide flashlamp. Larger exposed areas can be easily achieved by using longer flash tubes or multiple tubes, and smaller exposure areas are obtainable with an iris that more completely collimates the beam. The present invention provides a larger exposure area compared to prior art lasers or point sources and is very effective in the coagulation of blood vessels since blood flow interruption over a longer section of the vessel is more effective in coagulating it. The larger area exposed simultaneously also reduces the required procedure time.

Detector 22 (Figure 1) is mounted outside housing 12 and monitors the light reflected from the skin. Detector 22 combined with optical filters 18 and neutral density filters can be used to achieve a quick estimate of the spectral reflection and absorption coefficients of the skin. This may be carried out at a low energy density level prior to the application of the main treatment pulse. Measurement of the optical properties of the skin prior to the application of the main pulse is useful to determine optimal treatment conditions. As stated above, the wide spectrum of the light emitted from the non-laser type source enables investigation of the skin over a wide spectral range and choice of optimal treatment wavelengths.

In an alternative embodiment, detector 22 or a second detector system may be used for real-time temperature measurement of the skin during its exposure to the pulsed light source. This is useful for skin thermolysis applications with long pulses in which light is absorbed in the epidermis and dermis. When the external portion of the epidermis reaches too high a temperature, permanent scarring of the skin may result. Thus, the temperature of the skin should be measured. This can be realized using infra-red emission of the heated skin, to prevent over-exposure.

A typical real-time detector system would measure the infra-red emission of the skin at two specific wavelengths by using two detectors and filters. The ratio between the signals of the two detectors can be used to estimate the instantaneous skin temperature. The operation of the pulsed light source can be stopped if a preselected skin temperature is reached. This measurement is relatively easy since the temperature threshold for pulsed heating that may cause skin scarring is on the order of 50°C or more, which is easily measurable using infra-red emission.

The depth of heat penetration depends on the light absorption and scattering in the different layers of the skin and the thermal properties of the skin. Another important parameter is pulse-width. For a pulsed light source, the energy of which is absorbed in an infinitesimally thin layer, the depth of heat penetration (d) by thermal conductivity during the pulse can be written as shown in Equation 1:

$$d = 4 [k\Delta t/C\rho]^{1/2} \quad (\text{Eq. 1})$$

where

k = heat conductivity of the material being illuminated;
 Δt = the pulse-width of the light pulse;
 C = the heat capacity of the material;
 ρ = density of the material.

It is clear from Equation 1 that the depth of heat penetration can be controlled by the pulse-width of the light source. Thus, a variation of pulse-width in the range of 10^{-5} sec to 10^{-1} sec will result in a variation in the thermal penetration by a factor of 100.

Accordingly, the flashlamp 14 provides a pulse width of from 10^{-5} sec to 10^{-1} sec. For treatment of vascular disorders in which coagulation of blood vessels in the skin is the objective the pulse length is chosen to uniformly heat as much of the entire thickness of the vessel as possible to achieve efficient coagulation. Typical blood vessels that need to be treated in the skin have thicknesses in the range of 0.5mm. Thus, the optimal pulse-width, taking into account the thermal properties of blood, is on the order of 100msec. If shorter pulses are used, heat will still be conducted through the blood to cause coagulation, however, the instantaneous temperature of part of the blood in the vessel and surrounding tissue will be higher than the temperature required for coagulation and may cause unwanted damage.

For treatment of external skin disorders in which evaporation of the skin is the objective, a very short pulse-width is used to provide for very shallow thermal penetration of the skin. For example, a 10^{-5} sec pulse will penetrate (by thermal conductivity) a depth of the order of only 5 microns into the skin. Thus, only a thin layer of skin is heated, and a very high, instantaneous temperature is obtained so that the external mark on the skin is evaporated.

Figure 3 shows a variable pulse-width pulse forming circuit comprised of a plurality of individual pulse forming networks (PFN's) that create the variation in pulse-widths of flashlamp 14. The light pulse full width at half maximum (FWHM) of a flashlamp driven by a single element PFN with capacitance C and inductance L is approximately equal to:

$$\Delta t \approx 2[LC]^{1/2} \quad (\text{Eq. 2})$$

Flashlamp 14 may be driven by three different PFN's, as shown in Figure 3. The relay contacts R1', R2' and R3' are used to select among three capacitors C1, C2 and C3 that are charged by the high voltage power supply. Relays R1,

R2 and R3 are used to select the PFN that will be connected to flashlamp 14. The high voltage switches S1, S2 and S3 are used to discharge the energy stored in the capacitor of the PFN into flashlamp 14. In one embodiment L1, L2 and L3 have values of 100mH, 1mH and 5mH, respectively, and C1, C2 and C3 have values of 100mF, 1mF and 10mF, respectively.

In addition to the possibility of firing each PFN separately, which generates the basic variability in pulse-width, additional variation can be achieved by firing PFN's sequentially. If, for example, two PFN's having pulse-width Δt_1 and Δt_2 are fired, so that the second PFN is fired after the first pulse has decayed to half of its amplitude, then an effective light pulse-width of this operation of the system will be given by the relation: $\Delta t \approx \Delta t_1 + \Delta t_2$.

The charging power supply typically has a voltage range of 500V to 5kV. The relays should therefore be high voltage relays that can isolate these voltages reliably. The switches S are capable of carrying the current of flashlamp 14 and to isolate the reverse high voltage generated if the PFNs are sequentially fired. Solid-state switches, vacuum switches or gas switches can be used for this purpose.

A simmer power supply (not shown in Figure 3) may be used to keep the flashlamp in a low current conducting mode. Other configurations can be used to achieve pulse-width variation, such as the use of a single PFN and a crowbar switch, or use of a switch with closing and opening capabilities.

Typically, for operation of flashlamp 14 with an electrical pulse-width of 1 to 10msec, a linear electrical energy density input of 100 to 300J/cm can be used. An energy density of 30 to 100J/cm² can be achieved on the skin for a typical flashlamp bore diameter of 5mm. The use of a 500 to 650nm bandwidth transmits 20% of the incident energy. Thus, energy densities on the skin of 6 to 20J/cm² are achieved. The incorporation of the fluorescent material will further extend the output radiation in the desired range, enabling the same exposure of the skin with a lower energy input into flashlamp 14.

Pulsed laser skin treatment shows that energy densities in the range of 0.5 to 10J/cm² with pulse-widths in the range of 0.5msec are generally effective for treating vascular related skin disorders. This range of parameters falls in the range of operation of pulsed non-laser type light sources such as the linear flashlamp. A few steps of neutral density glass filters 18 can also be used to control the energy density on the skin.

For external disorders a typical pulse-width of 5 microsecond is used. A 20J/cm electrical energy density input into a 5mm bore flashlamp results in an energy density on the skin of 10J/cm². Cutting off the hard UV portion of the spectrum results in 90% energy transmission, or skin exposure to an energy density of close to 10 J/cm². This energy density is high enough to evaporate external marks on the skin.

Device 10 can be provided as two units: a lightweight unit held by a physician using handle 13, with the hand-held unit containing flashlamp 14, filters 18 and iris 20 that together control the spectrum and the size of the exposed area and the detectors that measure the reflectivity and the instantaneous skin temperature. The power supply, the PFN's and the electrical controls are contained in a separate box (not shown) that is connected to the hand-held unit via a flexible cable. This enables ease of operation and easy access to the areas of the skin that need to be treated.

The invention has thus far been described in conjunction with skin treatment. However, using a flashlamp rather than a laser in invasive treatments provides advantages as well. Procedures such as lithotripsy or removal of blood vessel blockage may be performed with a flashlamp. Such a device may be similar to that shown in Figures 1 and 2, and may use the electronics of Figure 3 to produce the flash. However, to properly couple the light to an optical fiber a number of couplers 40, 80 and 90 are shown in Figures 4 and 8-10, respectively.

Coupler 40 includes an optical source of high intensity incoherent and isotropic pulsed light such as a linear flash tube 42, a light reflector 44 which delivers the light energy to an optical fiber 46. The latter has a generally conical edge in the embodiment of Figure 4. Optical fiber 46 transfers the light from light collection system 44 to the treatment area. In general, coupler 40 couples pulsed light from a flash tube into an optical fiber and has applications in medical, industrial and domestic areas.

For example, coupler 40 may be used in material processing to rapidly heat or ablate a portion of a material being processed, or to induce a photo-chemical process. Alternatively, coupler 40 may be used in a photography application to provide a flash for picture taking. Using such a coupler would allow the flash bulb to be located inside the camera, with the light transmitted to outside the camera using an optical fiber. As one skilled in the art should recognize coupler 40 allows the use of incoherent light in many applications that coherent or incoherent light has been used in the past.

To provide for coupling the light to an optical fiber, flash tube 42 has a toroidal shape, shown in Figures 5 and 6, and is disposed inside reflector 44. In addition to the toroidal shape other shapes, such as a continuous helix, may be used for flash tube 42. However, a helical tube is more difficult to manufacture than a toroidal tube. Referring now to Figure 6, flash tube 42 is generally in the shape of a torus, but is not a perfect torus since the electrodes located at the end of the torus have to be connected to the power source. This does not create a significant disturbance in the circular shape of flash tube 42, since the connection to the electrodes can be made quite small.

Reflector 44 collects and concentrates the light, and has a cross-section of substantially an ellipse, in a plane perpendicular to the minor axis of the toroidal flash tube 42. The major axis of this ellipse preferably forms a small angle with the major axis of toroidal lamp 42. The exact value of the angle between the ellipse axis and the main axis of lamp 42 depends on the Numerical Aperture (NA) of the optical fiber. The toroidal flash tube is positioned so that its minor

axis coincides with the focus of the ellipse. The other focus of the ellipse is at the edge of optical fiber 46. Reflector 44 may be machined from metal with the inner surfaces polished for good reflectivity. Aluminum is a very good reflector with high reflectivity in the visible and ultraviolet wavelengths, and it may be used for this purpose. The reflector can be machined in one piece and then cut along a surface perpendicular to the main axis of the device. This will enable integration of the toroidal flash tube into the device.

As shown in Figure 4, the edge of optical fiber 46 is a cone with a small opening angle, so that the total area of the fiber exposed to the light from the flash tube is increased. Referring now to Figure 7 the geometry for coupling light into a conical tip is shown. It is assumed here that the light comes from a region in space with a refractive index of n_2 and that the conical section of the fiber (as well as the rest of the fiber core) has a refractive index of n_1 .

Not all the light rays hitting the cone are trapped in it. For light rays that propagate in a plane that contains the major axis of the system, a condition can be derived for the angle of a ray that will be trapped and absorbed in the fiber. This condition is shown in Equation 3.

$$\sin(\mu_{\text{criti}}) = \cos(\beta) \cdot [n_1^2/n_2^2 - 1]^{1/2} \sin(\beta) \quad (\text{Eq. 3})$$

Light will be trapped in the conical portion of the optical fiber if the incidence angle μ is larger than μ_{criti} calculated from Equation 3. Trapping is possible only if $n_1 > n_2$. If the medium outside of the fiber is air, $n_2 = 1$. Not all of the light trapped in the conical section of the fiber will also be trapped in the straight portion of the fiber if a fiber with a core and a cladding is used. If a fiber with a core and no cladding is used (air cladding), then all the rays captured in the conical section of the fiber will also be trapped in the straight section of the fiber.

The configuration shown in Figure 4 can also be used with a fluid filling the volume between the reflector and the optical fiber. A very convenient fluid for this purpose may be water. Water is also very effective in cooling the flashlamp if high repetition rate pulses are used. The presence of a fluid reduces the losses that are associated with glass to air transitions, such as the transition between the flashlamp envelope material and air. If a fluid is used in the reflector volume, then its refractive index can be chosen such that all the rays trapped in the conical section are also trapped in the fiber, even if core/cladding fibers are used.

Another way of configuring the fiber in the reflector is by using a fiber with a flat edge. This configuration is shown in Figure 8 and has trapping efficiency very close to the trapping efficiency of the conical edge. Many other shapes of the fiber edge, such as spherical shapes, can also be used. The configuration of the fiber edge also has an effect on the distribution of the light on the exit side of the fiber and it can be chosen in accordance with the specific application of the device.

The device may be used with a variety of optical fibers. Single, or a small number of millimeter or sub-millimeter diameter fibers, will typically be used in invasive medical applications. In other applications, particularly in industrial and domestic applications, it may be preferable to use a fiber having a larger diameter, or a larger bundle of fibers, or a light guide.

According to one embodiment flexible or rigid light guides are used to couple the light to the treatment area. Flexible light guides made from a bundle of quartz or other glass fibers that are fused together by heat at the edge of the bundles. The bundles may be circular, rectangular, or any other useful shape. Rigid light guides may be made from quartz, acrylic, glass, or other materials having a high degree of transparency. The material is generally highly polished on all sides.

A typical cross section of a circular light guide useful for therapeutic treatment is one mm to ten mm in diameter. Alternatively, a rectangular light guide may be used having typical dimensions of 3 mm by 10 mm to 30 mm by 100 mm. In either case the length may be 20 to 300 mm, or as needed for the specific application.

According to another alternative embodiment a rectangular light guide is used to more efficiently couple the light. The rectangular light guide is chosen to have a shape that matches a rectangular linear flashlamp and to match the shape of the vessel being treated.

The light guides described above may be used in another alternative embodiment to control the spectrum of light delivered to the treatment area. Spectral control can be achieved by making the light guide from a material that had an absorbing dye dissolved therein. Thus, light transmitted by the light guide will have a spectrum in as determined by the absorbing dye. Alternatively, a flat, discrete filter may be added to one end (preferably the input end) of the light guide. Both of these filters are absorbing filters. The inventors have found that absorbing filters produced by Schott, having Model Nos. OG515, OG550, OG570, and OG590 have suitable characteristics.

Additionally, interference filters or reflective coatings on the light guide may be used by applying a proper optical coating to the light guide. Again, a single discrete interference filter could also be used. Additionally, combinations of the various filters described herein, or other filters, may be used. The use of the filters described here may render the use of the filters described earlier with reference to Figure 1 redundant.

An alternative embodiment entails the use of application specific light guides. In this way the spectra of light for various treatments can be easily controlled. According to this alternative each type of treatment will be performed with a specific light guide.

The optical properties of the light guide will be chosen to optimize the particular treatment. The wavelengths below are particularly useful for the respective treatments:

arteries less than 0.1 mm in diameter - 520-650nm

veins less than 0.1 mm in diameter - 520-700nm

5 vessels between 0.1 and 1.0 mm in diameter - 550-1000nm

larger vessels - 600-1000nm

In each case if the skin is darker (higher pigmentation) longer wavelengths on the lower cut-off portion of the spectrum should be used.

10 Multiple spectra may be used for optimal penetration. This may be accomplished by illuminating with a few pulses, each having a different spectrum. For example, the first pulse can have a spectrum that is highly absorbed in blood. This pulse will coagulate the blood, thereby changing the optical properties of the blood, making it more absorbing in another wavelength range (preferably longer). A second pulse will be more efficiently absorbed since the blood absorbs energy of a greater wavelength range. This principle may be used with lasers or other light sources as well.

15 In addition to the features of the light guides discussed above, a light guide is used, in one alternative embodiment, to control the angular distribution of the light rays impinging on the skin. Light that impinges on the skin at large angles (relative to the perpendicular) will not penetrate very deeply into the tissue. Conversely, light that impinges perpendicularly to the skin will have a deeper penetration. Thus, it is desirable to provide a distribution of light rays that has a relatively wide angular divergence when the treatment requires shallow penetration. Alternatively, a narrow divergence is preferable for treatment requiring deep penetration is desired. Some treatment might require both shallow and deep penetration.

20 Figure 15 shows a light guide 115 having an exit beam with a greater angular divergence than that of the entrance beam. As shown in Figure 15, a beam 116 enters light guide 115 at a small angle, relative to the axis of light guide 115. When beam 501 exits light guide 115, the angle, relative to the axis, is much greater. The tapered shape of light guide 115 enhances this divergence.

25 Figure 16 shows a straight light guide 118 that maintains the angular distribution of the rays of light that enter into it. A beam 119 is shown entering and exiting light guide 118 with the same angle, relative to the axis of coupler 601. Alternate use of both light guides 115 and 118 can achieve the narrow and deep penetration discussed above. Alternatively, the user can select the type of coupler according to the depth of penetration needed for the treatment being performed.

30 Figures 9 and 10 show a coupler 90 for coupling a linear flash tube 92 through a linear to circular fiber transfer unit 94 to a fiber bundle 96. A reflector 98 has an elliptical cross-section, shown in Figure 10, in a plane parallel to the axis of linear flash tube 92 in this embodiment. Tube 92 is located on one focus of the ellipse while the linear side of linear to circular bundle converter 94 is located at the other focus of the ellipse. This configuration is relatively simple to manufacture and commercially available linear to circular converters such as 25-004-4 available from General Fiber Optics may be used. This configuration is particularly useful for larger exposure areas of the fiber, or for flash illumination purposes.

35 The energy and power densities that can be achieved by this invention are high enough to get the desired effects in surface treatment or medical applications. For the embodiment shown in Figure 4 the total energy and power densities can be estimated as follows. For a typical toroidal lamp with a 4mm bore diameter and a major diameter of 3.3cm an electrical linear energy density input of 10J/cm into the lamp can be used with a 5μsec pulse width. The light output of the lamp will be 5 to 6J/cm for optimal electrical operating conditions. For the reflector shown in Figure 4, 50% of the light generated in the lamp will reach the lower focus. Thus, a total energy flux on the focus of 25 to 30J may be obtained. For embodiments shown in Figure 4 or Figure 8 the total cross-section area of reflector at the focal plane has a cross-section of 0.8cm². Energy densities on the order of 30 to 40J/cm² at the entrance to the fiber should be attained with this cross-section. This corresponds to power densities of 5 to 10MW/cm², which are the typical power densities used in medical or material processing applications.

40 For longer pulses, higher linear electrical energy densities into the lamp can be used. For a 1msec pulse to the flash tube a linear electrical energy density of 100J/cm can be used. The corresponding energy density at the focal area would be up to 300J/cm². Such energy densities are very effective in industrial cleaning and processing applications as well as in medical applications.

45 Alternative embodiments for coupling the optical fiber to an extended light source such as a linear flashlamp are shown in Figures 11 and 12. In the embodiment of Figure 11 an optical fiber 101 is wound around a lamp 102 and a lamp envelope 103. Some of the light that is produced by the light source is coupled into the fiber. If the light rays are propagating in the direction that is trapped by the fiber then this light will propagate in the fiber and it can be used at a fiber output 104. One limitation of this configuration is the fact that most of the light emitted by lamp 103 travels in a direction perpendicular to the surface of lamp 103 and cannot be trapped in fiber 101.

50 The embodiment shown in Figure 12 overcomes this problem. A doped optical fiber 105 is wound around lamp 102 and envelope 103, rather than an undoped fiber such as fiber 101 of Figure 11. The dopant is a fluorescent material which is excited by the radiation emanating from lamp 102 and radiates light inside the fiber. This light is radiated omni-

directionally and the part of it that is within the critical angle of fiber 105 is trapped and propagates through the fiber and can be used at fiber output 104. The angle of light that is trapped in the fiber is the critical angle of the material from which the optical fiber or optical wave guide is made. For a fiber (or optical wave guide) in air this angle is given by $\sin \alpha = 1/n$.

Typically for glass or other transparent materials $n = 1.5$ and $\alpha = 41.8^\circ$. This corresponds to a trapping efficiency of more than 10% of the light emitted by fluorescence inside the fiber. If we assume a 50% efficiency of the fluorescence process we find out that more than 5% of the light produced by the lamp is trapped and propagated down the fiber. For example, a 4" lamp with a linear electrical energy input of 300J/inch and 50% electrical to light conversion efficiency would couple 2.5% of its electrical energy into the fiber. This corresponds, for the 4" lamp case to a total light energy of 30J of light. This embodiment has the additional advantage of transferring the wavelength emitted by the lamp to a wavelength that may be more useful in some of the therapeutic or processing applications mentioned before. Thus, fluorescent material doped in the fiber can be chosen in accordance with an emission wavelength determined by the specific application of the device.

One alternative embodiment includes the use of a gel to couple the light to the skin. This alternative reduces heating of the outer layer of the skin (the epidermis and upper layers of the dermis). The gel is preferably a high viscosity water based gel and is applied to the skin before treatment, although other gels that are not necessarily water based may be used. A gel having a relatively high heat capacity and thermal conductivity, such as a water based gel, is preferable to enable cooling of the outer skin (the epidermis in particular). Transparency is also desirable because during treatment light passes through the transparent gel and reaches the skin.

Referring now to Figure 13, a gel 110 is applied to the skin 21 prior to the treatment. A flat layer of gel on top of the skin is used since irregularities in the upper layer of the gel through which the light passes may cause scattering of the light and reduce its penetration into the skin. In order to achieve this flatness a solid, transparent, flat piece 111 may be applied on top of the skin. The configuration is shown schematically in Figure 13. The transparent plate can be made out of glass or other transparent materials. Either the flashlamp housing or the light guides discussed above may be placed in direct contact with the transparent plate.

The configuration of Figure 13 has the advantage of reducing the scattering of light (represented by arrows 113) that enters into the skin due to irregularities in the surface of the skin. The skin has an index of refraction that is larger than that of the air. As a result, any photon that impinges on the air skin interface is deflected if it does not hit the skin at an incidence angle of 0° . Since the surface of the skin is irregular the angular distribution of the skin increases. This is shown schematically in Figure 14.

The use of gel addresses this problem since the gel can fill irregular voids that are created by the skin structure. The transparent plate that covers the gel and the gel itself will preferably have an index of refraction that is close to that of the skin. This is relatively easy since the index of refraction of the skin is of the order of 1.4 in the visible and the near infrared. Most glasses and transparent plastics have indices of refraction that are of the order of 1.5 which is close enough. The index of refraction of water is of the order of 1.34 in this range. Water based gels will have similar indices of refraction. The index can be increased by proper additives. The plate and gel thus act as a flat surface for the light to impinge upon. Because the gel and plate have an index of refraction close to that of the skin there is very little scattering at the gel-plate and gel-skin interfaces.

The use of a gel has been experimentally successful in the treatment of leg veins and other benign vascular lesions of the skin. The treatments were carried out with the flashlamp described above. However, in alternative embodiments a different incoherent source, or a coherent source, may be used.

During operation light is typically applied to the skin in a sequence of three pulses with short delays between the pulses. This mode of operation is used in order to take advantage of the faster cooling of the superficial, thin (less than 0.1mm thick) epidermis compared to the larger and deeper vessels typical of leg veins. The gel in contact with the skin cools the epidermis during the waiting period between the pulses. This cooling reduces significantly the damage to the epidermis.

In accordance with the invention, light is applied to the treated area in either a long pulse or in a sequence of pulses separated by a delay. The delay and/or pulse length is preferably controlled by the operator to provide enough heat to accomplish the desired treatment but not enough heat to damage the skin.

This concept was tested with large and deep vessels (of the order of 2mm in diameter and 2mm deep). A thin layer of commercial water based ultrasound gel (1 to 2mm thick, "Aqua clear" gel made by Parker USA) was applied on the skin. A 1mm thin glass window was used to generate a flat layer of the gel. The light from the device passed through the thin glass and the gel and into the skin. Care was taken to assure that no air bubbles exist in the gel. This configuration was tested with photon fluences of 30 to 50J/cm². Coagulation and clearance of the vessels was obtained without causing damage to the skin. This is contrary to similar trials in which gel was not used and in which fluences of 20J/cm² with the same pulse structure caused burns of the skin.

The epidermis has a thickness of approximately 0.1 mm and a cooling time of about 5 msec. Thus, to avoid burning delays greater than 5 msec are used.

In another alternative embodiment the spectrum of the light used for treatment is controlled by controlling the voltage and/or current applied to the flashlamp. As is well known in the art, the spectrum of light produced by a flashlamp is dependent on the voltage and current provided to the flashlamp. According to this embodiment the input voltage and current is selected to provide a desired treatment spectrum. The appropriate voltage and currents may be determined experimentally for each flashlamp used. For example, a flashlamp current of 200 amps produced the spectra shown in Figure 17. Similarly, the spectra of Figure 18 was produced using a flashlamp current of 380 amps. The spectra of Figure 17 shows a significant enhancement in the wavelength range of 800-1000nm. Such a spectra is particularly useful for treatment of large vessels.

The different currents and voltages used to control the output spectra may be obtained using a group or bank of capacitors that are capable of being connected in either series or parallel as part of the power source for the flashlamp. A series connection will provide a relatively high voltage and high current, thereby producing a spectra having energy in a shorter wavelength, such as 500-650 nm. Such a series connection will be more appropriate for generating shorter pulses (1 to 10 msec, e.g.) useful for treatment of smaller vessels.

A parallel connection provides a lower current and voltage, and thus produces an output spectra of a longer wavelength, such as 700-1000 nm. Such a spectra is more appropriate for treatment of larger vessels and is suitable for producing longer pulses (in the range of 10-50 msec, e.g.). The selection of series or parallel connections may be done using a relay or sets of relays.

In one alternative embodiment the pulse forming network of Figure 3 is replaced by a GTO driver circuit 121, such as that shown in Figure 19. The driver circuit of Figure 19 uses a switch capable of being turned both on and off to control the application of power to the flashlamp. While this alternative embodiment will be described with respect to a GTO being used as the switch, other switches capable of being turned both on and off, such as IGBTs, may also be used.

Referring now to Figure 19, driver circuit 121 includes a high voltage source 122, a capacitor bank C5, an inductor L5, a diode D5, a switch GTO1, a diode D6, a diode D7, a resistor R5, a capacitor C6, a GTO trigger generator TR1, a resistor R7, a capacitor C7 and a flashtube trigger generator TR2. These components are connected to flashlamp 14 and serve to provide the power pulses to flashlamp 14. The duration and timing of the pulses are provided in accordance with the description herein. Driver 121 operates in the manner described below.

High voltage source 122 is connected across capacitor bank C5, and charges capacitor bank C5 to a voltage suitable for application to flashlamp 14. Capacitor bank C5 may be comprised of one or more capacitors, and may be configured in the manner described above.

Prior to illumination of flashlamp 14 flashtube trigger generator TR2 breaks down flashlamp 14 and creates a relatively low impedance channel therein. After the flashlamp breaks down, capacitor C7 dumps current into flashlamp 14, further creating a low impedance channel in flashlamp 14. In this manner a pre-discharge is provided that prepares flashlamp 14 for the power pulse. Capacitor C7 provides a small amount of current, relative to capacitor bank C5. Alternatively, driver circuit 121 may operate in a simmer mode, wherein the pre-discharge is not necessary.

Thereafter, switch GTO1 is turned on via a pulse from GTO trigger generator TR1, completing the circuit between flashlamp 14 and capacitor bank C5. Thus, capacitor bank C5 discharges through flashlamp 14. An inductor L5 may be provided to control the rise time of the current through flashlamp 14. Inductor L5 may include an inherent resistive component, not shown.

After a length of time determined by the desired pulse width has passed, GTO trigger generator TR1 provides a pulse to switch GTO1, turning it off. A control circuit determines the timing of the trigger pulses and provides them in accordance with the desired pulse widths and delays.

A snubber circuit comprised of diode D6, resistor R5, and a capacitor C6 is provided for switch GTO1. Also, diodes D5 and D7 are provided to protect switch GTO1 from reverse voltages. Resistor R7 is provided in parallel with flashlamp 14 to measure the leakage current of switch GTO1, which can in turn be used to make sure that switch GTO1 is operating properly.

A possible addition to driver circuit 121 is to provide an SCR or other switch in parallel with capacitor bank C5. This allows the discharge or resetting of capacitor bank C5 without turning on switch GTO1. Other modifications may be made, such as providing the circuit with a serial trigger, rather than the parallel trigger shown. Another modification is to use the driver circuit with a laser rather than flashlamp 14.

Proper use of pulse widths and delays can aid in avoiding burning the epidermis. The epidermis has a cooling time of about 5 msec, while large vessels have a longer cooling time (a 1 mm vessel has a cooling time of about 300 msec). Thus, during a pulse of duration longer than 5 msec the epidermis can cool down but the vessel will not. For example, for treatment of a large vessel (such as one having a diameter of about 1mm) a pulse of 100 msec will allow the skin to cool, but the vessel will not cool.

The same effect may be achieved using trains of pulses. This is useful when it is not practical to provide a single long pulse to the flashlamp. The delays between pulses are selected to allow the skin to cool, but to be too short for the vessel to cool. Thus, larger vessels can be treated with longer delays because they have greater cooling times. Small vessels cool quickly and long delays are not effective. However, they also need less energy and can be treated effectively in a single pulse.

Typical delay times are in the range of 20 msec to 500 msec. More specifically, delays of between 100-500 msec are effective for vessels larger than 1 mm in diameter. Delays of between 20-100 msec are effective for vessels between 0.5 and 1 mm in diameter. Delays of between 10-50 msec are effective for vessels between 0.1 and 0.5 mm in diameter. A single pulse having a width in the range of 1 msec to 20 msec is effective for vessels less than 0.1 mm diameter.

5 Additionally, delays should be selected according to skin pigmentation. Darker skin absorbs more energy and needs more time to cool: thus longer delays are needed. Lighter skin absorbs less energy and can accommodate shorter delays.

It has been found that multiple pulses avoids "purpura" or the explosion of small vessels in or close to the skin. The use of pulses to avoid burning and provide cooling will be effective for light provided by lasers or other sources as well.

10 Another alternative embodiment includes the use of a microprocessor or personal computer to control the flash-lamp. The microprocessor can be used to provide the timing functions and prompt the trigger signals described above. Additionally, in one embodiment the microprocessor includes a user interface, such as a screen and keyboard, buttons, mouse, or other input device. The microprocessors have information stored therein that aids in the selection of treatment parameters.

15 For example, if the condition being treated is a port wine stains skin type III, the physician inputs that condition into the microprocessor. The microprocessor responds with suggested treatment parameters, such as using a 570nm cut-off filter, a double pulse with a delay of 50 msec and a fluence of 55 J/cm². The physician can alter these suggested parameters, but need not refer back to operating guidelines for suggested parameters.

The microprocessor or personal computer can also be used to create and store patient information in a database. 20 Thus, past treatment information such as condition being treated, treatment parameters, number of treatments, etc. is stored and may be recalled when the patient is again treated. This aids in providing the proper treatment to the patient. Additionally, the database may include photographs of the patient's condition before and after each treatment. Again, this aids in record keeping and determining what treatments are most successful for given conditions.

25 In addition to the treatments described above the devices and methods described herein may be used to treat other conditions. For example, psoriasis and warts have been successfully treated. Similarly, skin rejuvenation (treating wrinkles) should be effective. The inventor further contemplates using this invention to treat hemorrhoids, throat lesions, and gynecological problems associated with vascular malformations.

30 Thus, it should be apparent that there has been provided in accordance with the present invention a flashlamp and coupler that fully satisfy the objectives and advantages set forth above. Although the invention has been described in conjunction with specific embodiments thereof, it is evident that many alternatives, modifications and variations will be apparent to those skilled in the art. Accordingly, it is intended to embrace all such alternatives, modifications and variations that fall within the spirit and broad scope of the appended claims.

Claims

35 1. A therapeutic treatment device for treating a treatment region comprising an incoherent, pulsed, light source operable to provide a light output for treatment, a power supply connected to the light source, a housing including a reflector and having an opening, wherein the light source is disposed within the housing and the reflector reflects light from the light source to the opening, and a flexible light guide is disposed between the opening and the treatment region, wherein the light guide receives the incoherent light from the light source and transmits the light to the treatment region and the light source, reflector and light guide cooperate to provide between 6 and 100 J/cm² to the skin, characterized in that:

the light guide transmits light having a predetermined angular divergence, wherein the divergence is selected in response to a desired treatment depth.

45 2. The treatment device of claim 1 further characterized in that:
a first interference filter is disposed between the light source and the light guide; and
a second absorbing filter is disposed between the first filter and the treatment region.

50 3. The treatment device of claim 3 further characterized in that:
the light guide is made from a material including an absorbing dye and is the second absorbing filter.

4. The treatment device of claim 1 further characterized in that the reflector includes a reflecting portion that is a portion of a circle.

55 5. The treatment device of claim 4 further characterized in that the reflecting portion is relatively close to the light source.

6. The treatment device of claim 5 further characterized in that a cooling gel is disposed over the treatment area.

7. A method for therapeutically treating a treatment region comprising the steps of producing incoherent, pulsed, light in a housing, reflecting the light to an opening in the housing, transmitting the light from the opening to the treatment area through a light guide to provide between 6 and 100 J/cm₂ to the skin, characterized in that:
the light guide transmits light such that the light has a predetermined angular divergence and the divergence is selected in response to a desired treatment depth.
8. The treatment method of claim 7 further characterized in that:
the light is filtered by a first interference filter, between the light source and the light guide; and
the light is filtered by a second absorbing filter between the first filter and the treatment region.
9. The treatment method of claim 8 further characterized in that:
the light is filtered by the second absorbing filter as the light passes through the light guide.
10. The method device of claim 7 further characterized in that the step of reflecting includes the step of reflecting the light by a reflector that a portion of a reflecting surface is a portion of a circle.
11. The treatment method of claim 10 further characterized in that the reflecting portion is relatively close to the light source.
12. The treatment method of claim 7 further including the step of applying a cooling gel to the treatment area.

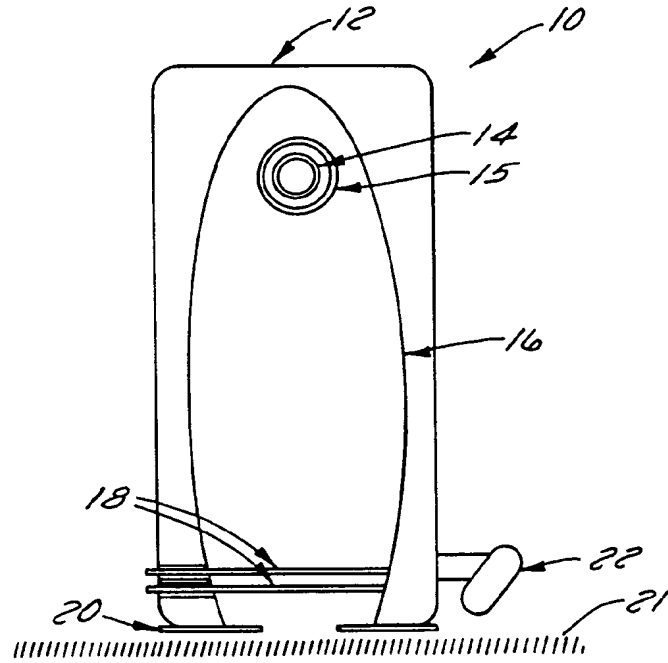


FIG. 1

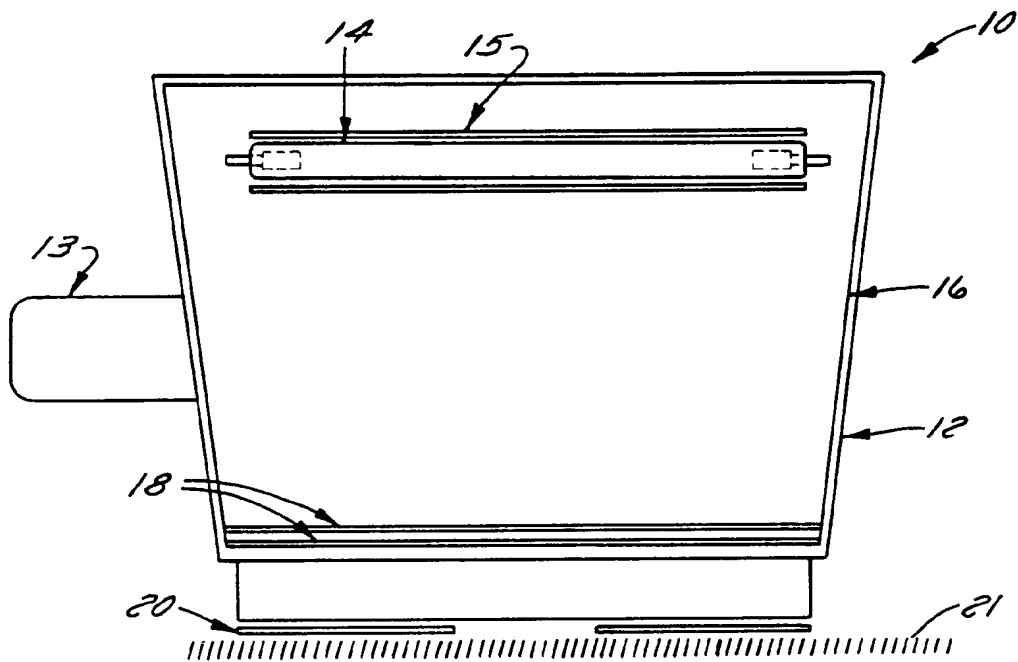


FIG. 2

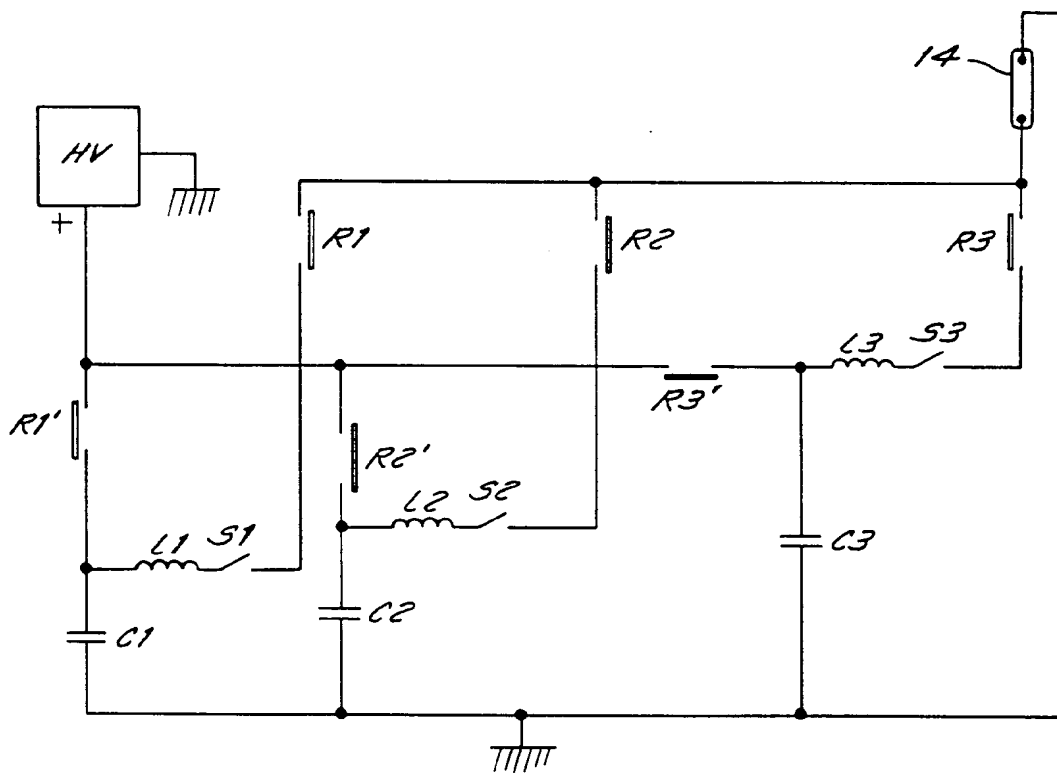


FIG. 3

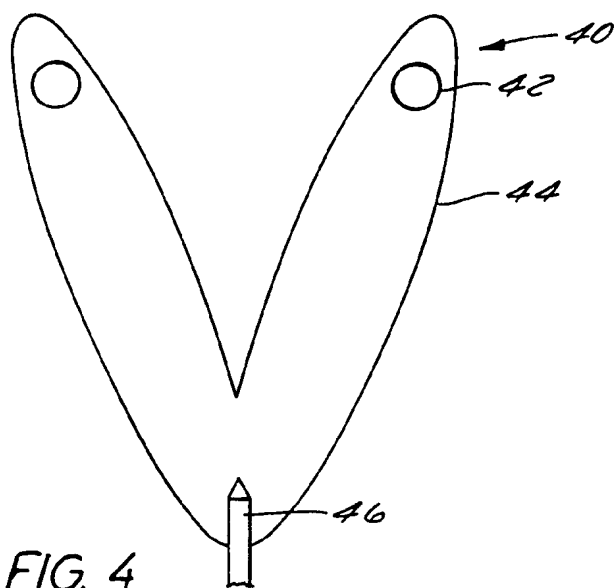


FIG. 4

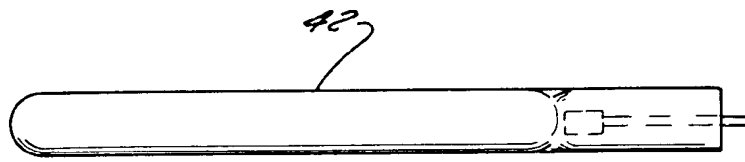


FIG. 5

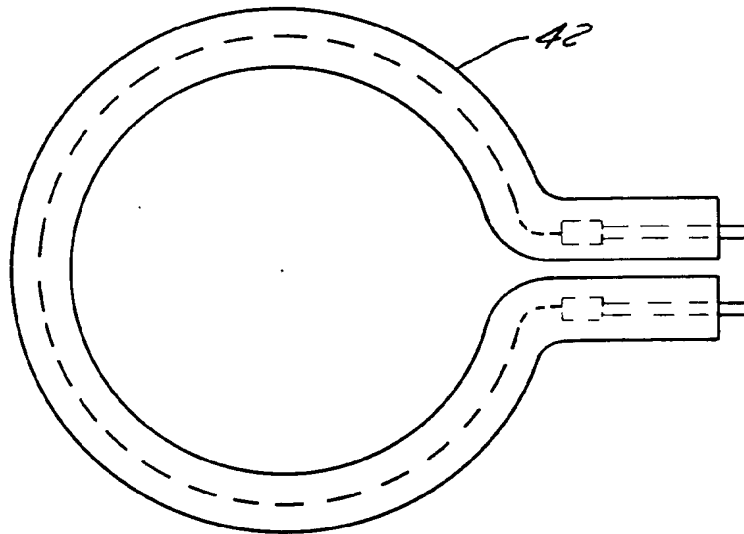


FIG. 6

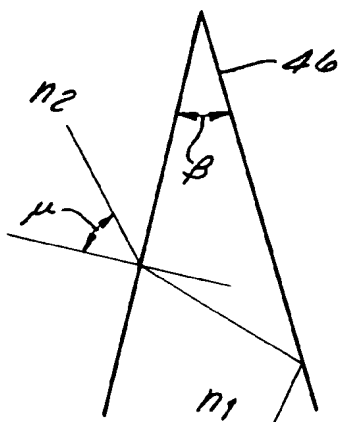


FIG. 7

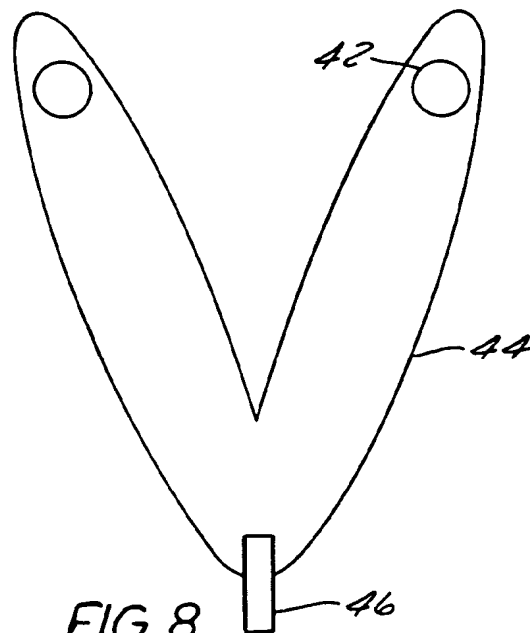


FIG. 8

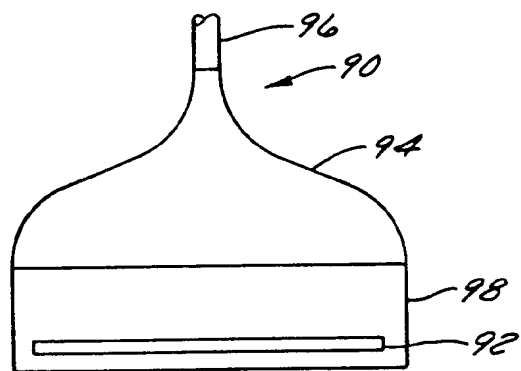


FIG. 9

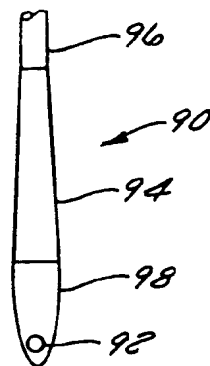


FIG. 10

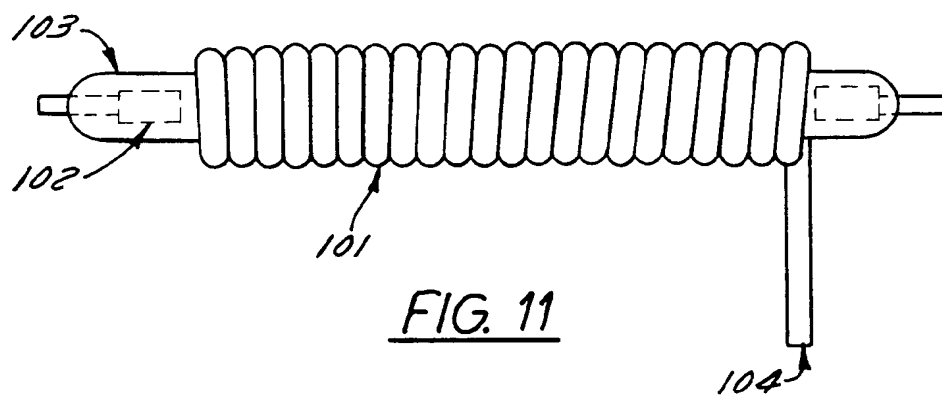


FIG. 11

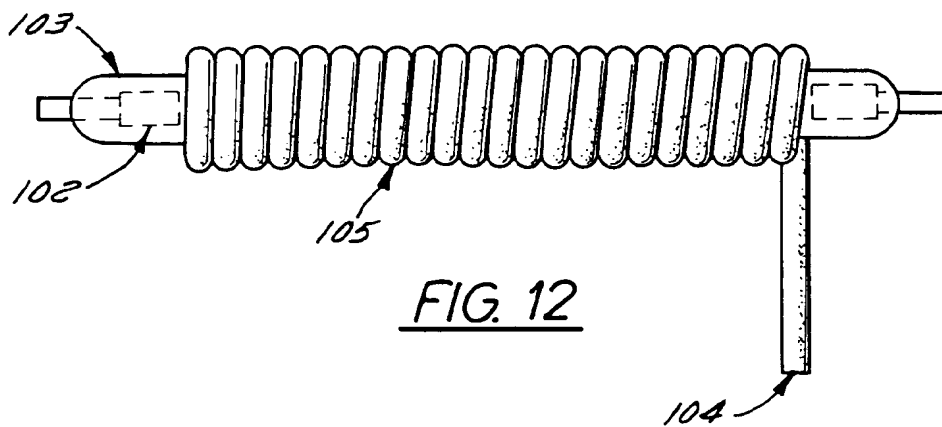


FIG. 12

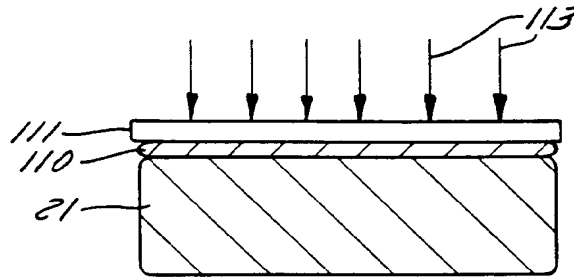


FIG. 13

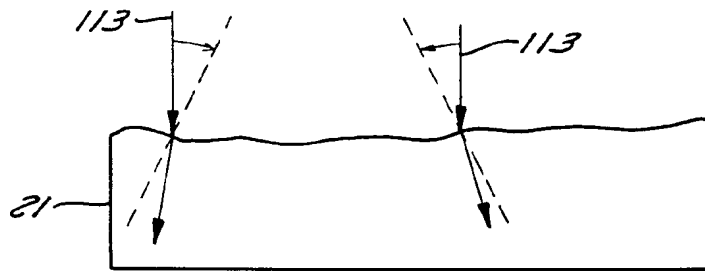


FIG. 14

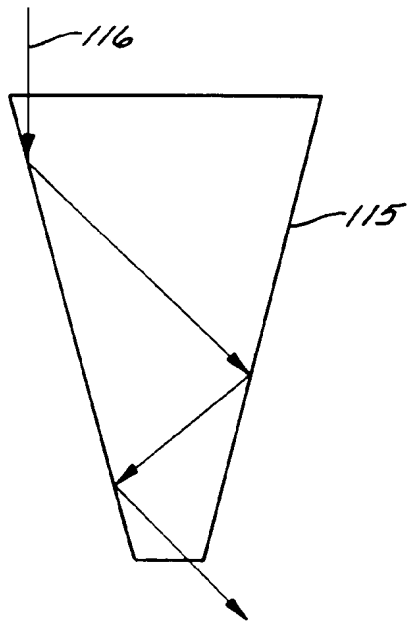


FIG. 15

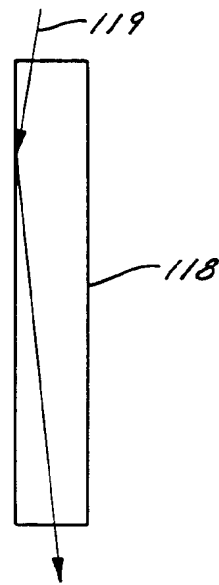
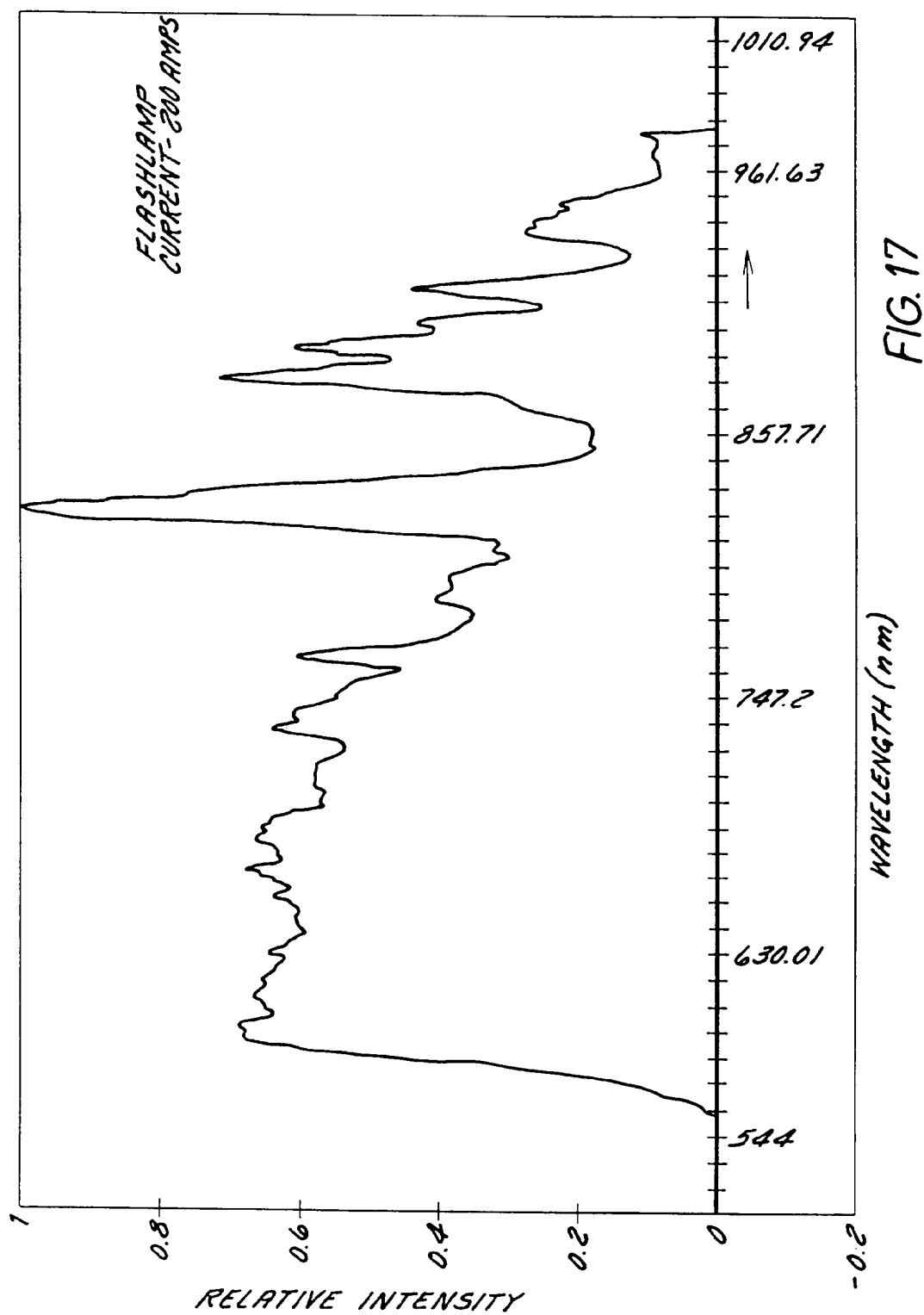
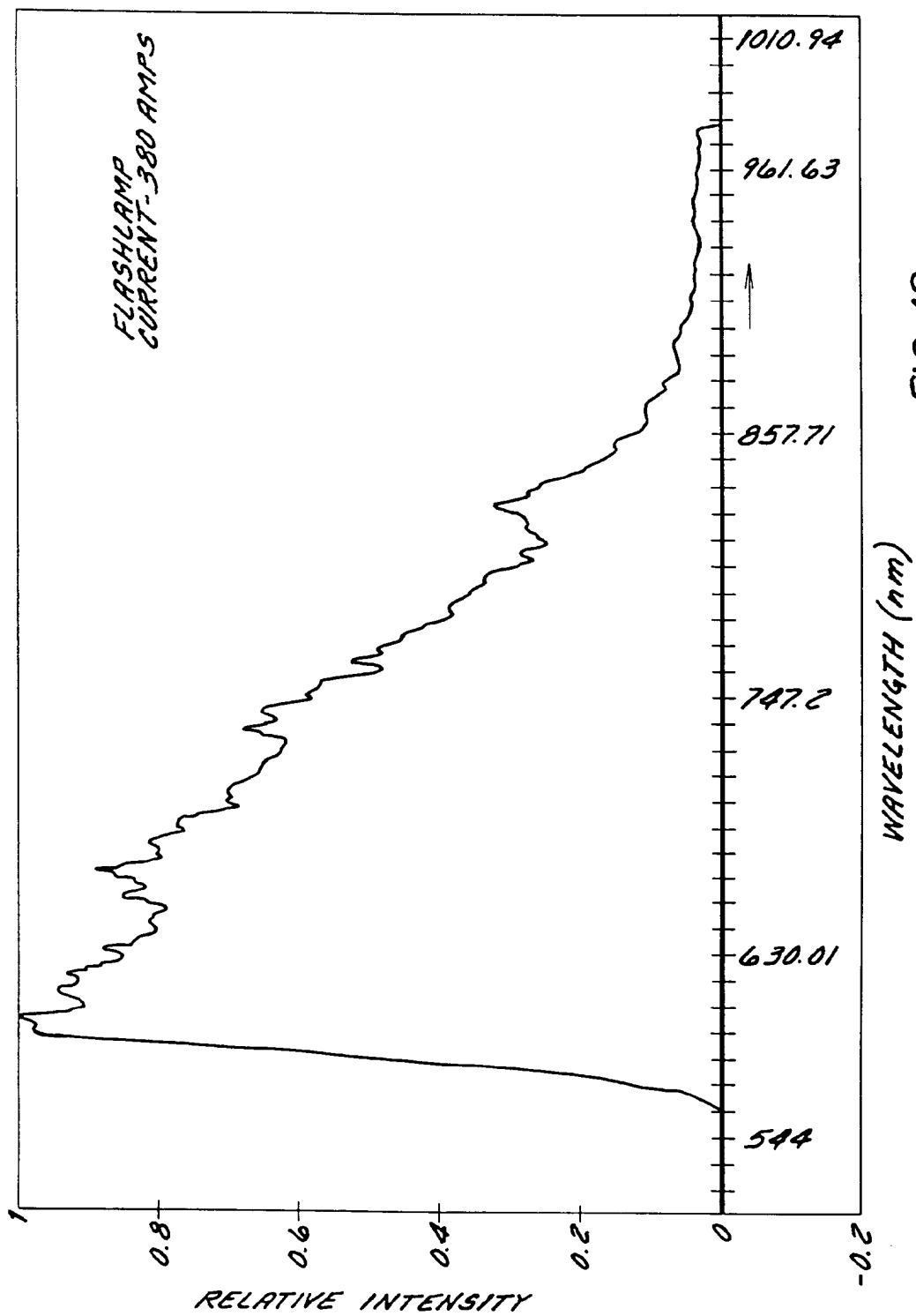
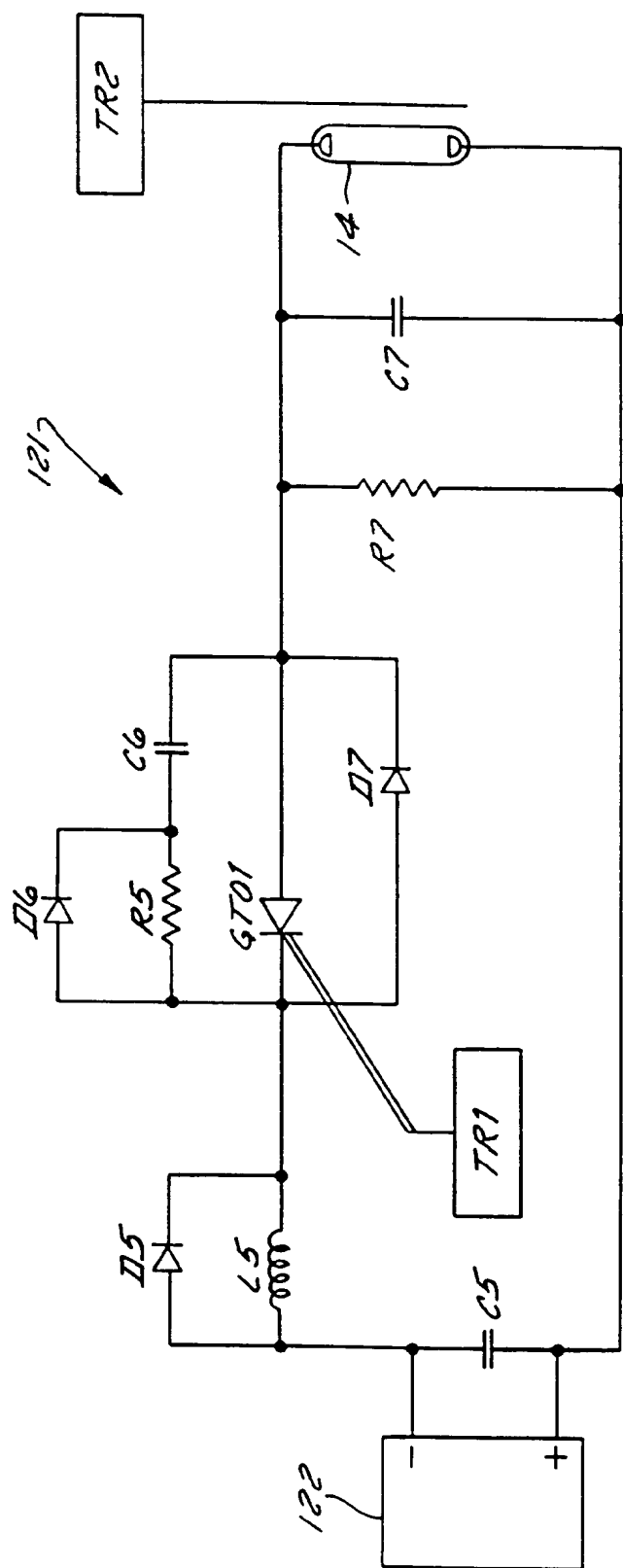


FIG. 16



FIG. 18

FIG. 19



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(54) **Method and apparatus for the diagnostic and composite pulsed heating and photodynamic therapy treatment**

(57) The present invention describes an apparatus for therapeutically treating cancerous cells within a tissue having a plurality of vessels for supplying blood thereto. The apparatus comprises a housing (100), including a window through which light can pass; a source of light (102, 112), capable of being operated in both a

continuous mode and a pulsed mode, disposed in the housing; and a reflector (104, 114), disposed in the housing, that directs the light through the window. The pulsed light source may be a noncoherent light. A filter system (106, 108) may be disposed in the path of the light to limit the light to a range of wavelengths.

Description

The present invention is directed toward an apparatus and method for diagnostic and composite treatment of a wide range of solid tumors. More particularly, the invention relates to an efficient apparatus and method for diagnosing and simultaneously treating certain types of cancer with photodynamic therapy and pulse heating.

In general, the fundamental mechanism of photodynamic therapy (PDT) action is initiated by the absorption of visible light by a tumor that has been injected with a photosensitizing agent. As a result of the reaction of incident light in the range of 600 to 1000 nanometers (nm) with the photosensitizing agent, oxygen is generated in the singlet state. The reaction of this photochemically generated singlet oxygen with the intracellular lipids, proteins, and nucleotides harms the cancerous cells and ultimately results in tumor necrosis.

In the prior art, lasers and continuous wave (CW) noncoherent light sources were used for PDT. The delivery dose of the light energy varied in the range of 25 to 200 joules per square centimeter (J/cm^2) with a fluency rate of 10 to 200 milliwatts per square centimeter (mW/cm^2), depending upon the tumor size and the spectrum of the radiation.

Typically, the light source used for clinical PDT of solid tumors is a CW tunable dye laser that is pumped by an argon ion laser. Alternatively, a frequency doubled Nd:YAG pumped dye laser which produces pulsed light may be used.

As an alternative to laser light sources, CW wavelength filtered lamp sources may be used. For example, the use of a xenon short arc lamp of 150 W possessing a narrow light beam with a spectral region in the range of 610 to 750 nm is claimed in United States Patent No. 5,344,434, issued on September 6, 1994, to Eli T. Talmore, entitled "Apparatus For The Photodynamic Therapy Treatment." A glass lens focuses the light beam to a range of 3 to 12 millimeters (mm) in diameter. The light beam is then delivered to the target through a light guide. To obtain the desired dose of radiation with this device, the treatment time preferably ranges from approximately 20 to 60 minutes, depending on the size of the tumor.

In the prior art, one cancer diagnostic method that is used is based on the fluorescence of malignant cells in the wavelength range of 400 to 750 nm under illumination of near ultraviolet (UV) and blue radiation in the range of 300 to 400 nm. The presence of malignant cells reduces the autofluorescent intensity in the blue green wavelength region, thus providing a signal that is distinguishable from the higher intensity signal originating from surrounding healthy tissue. Alternatively, diagnostic techniques may employ different types of injected photoactivators. Accumulation in the tumor of these alternate types of photoactivators results in increased fluorescence of the malignant cells in comparison to the

surrounding healthy tissue.

Only a few types of light sources that can excite tissue autofluorescence are currently available. One example of such a light source is a nitrogen laser which emits 3 nanosecond (nsec) light pulses having a 337 nm wavelength. Another type of light source is a UV source, such as an excimer pumped dye laser, that produces a 308 nm light beam which is focused into a 600 micrometer (μm) thick optical fiber. As an alternative to the use of lasers as the exciting light source, mercury lamp sources which filter two excitation wavelengths of 365 and 405 nm may be employed.

There is, however, a need for a simple tunable apparatus and method for providing efficient PDT treatment for a wide range of tumor parameters, including size of the tumor and depth of location. Such an apparatus and method will preferably be able to control the fluency rate and spectrum of output radiation, dependent on the type of photosensitizing agent used, to achieve efficient PDT treatment. In addition, the apparatus and method preferably will produce radiation with the appropriate wavelength range desirable for cancer diagnostics. In addition, an apparatus and method using pulsed light rather than CW light would be advantageous because pulsed light enables better temperature control and the achievement of important hyperthermia effects.

The present invention achieves efficient diagnosis and treatment of cancerous cells. The method and apparatus of the invention simultaneously provide photodynamic therapy and pulsed heating of the tumor, which accelerate the photochemical reaction and coagulate the blood, thus limiting the supply of blood to the tumor.

The apparatus of the present invention includes either a linear flashlamp in a housing with a straight reflector, or a bent flashlamp with a cone reflector, or any other light source suitable for the desired use. When used for diagnostic purposes, interchangeable interference filters provide a spectral region of 350 to 500 nm with a peak at 400 nm. The apparatus also includes an interchangeable filter which filters radiation with wavelengths shorter than 520 nm. Another interchangeable filter provides either a spectral region of 600 to 700 nm or a spectral region of above 600 nm. The produced light beam may be directed to the target through either a flexible light guide or a shorter quartz light guide, or directly from the output of the light source.

In the diagnostic mode, the invention can also be operated in either a pulsed or CW mode. In the pulsed mode a single pulse or repetitive pulses having a frequency of 0.02 to 2 pulses/sec may be used. Pulse duration may be varied over a range, for example from 0.1 to 100 msec. Optical energy density per pulse ranging from 0.02 to 4 J/cm^2 (in each pulse) and an illumination area that varies in size dependent upon the distance of the target area from the light guide may also be provided.

In the CW mode the suspected cancerous area is

continuously illuminated while the physician looks for fluorescence that indicates the presence of a tumor. Illumination can be carried out through a light guide or by directly exposing the area to the opening in the housing that contains the lamp. The light guide is particularly useful for illuminating internal objects and areas that are difficult to access. When treating large surface areas, the light source may be used without the light guide.

Similarly, in the PDT treatment mode either pulses or a CW may be provided.

Pulses having a frequency ranging from 0.1 to 1 pulse/sec and durations of from 0.1 to 100 msec may be used. The output spectrum is preferably in the range of either 600 to 1000 nm or 600 to 700 nm. The light beam may be delivered to the target either directly or by a flexible light guide if internal treatment is desired and for external treatment of a large surface, the light beam can be delivered directly to the target without a light guide. An optical energy density per pulse ranging from 0.1 to 20 J/cm² (pulse) in repetitive mode is preferably provided as is a fluency rate ranging from 100 to 2000 mW/cm². The illumination area preferably varies from 0.5 to 3 cm².

In the CW treatment mode the treated object is continuously illuminated by the filtered lamp radiation. Illumination can be achieved either through a light guide or by directly exposing the treated area to the opening in the housing that contains the lamp. The light guide can be used for treating internal and small size tumors, and areas that are difficult to access. When treating large surface areas, the light source may be used without the light guide.

The apparatus and method of the invention can be used for photodynamic therapy treatment using a variety of different photosensitizers. The resulting irradiance from the invention is significantly higher than the radiation produced by the commonly used laser and noncoherent light sources. In addition, the invention permits treatment of large areas in a much shorter time than is possible with currently used methods. Also, the invention provides a further advantage in that it is very safe and poses minimal risk of accidental harm to the operator and the patient.

Other principal features and advantages of the invention will become apparent to those skilled in the art upon review of the following drawings, the detailed description, and the appended claims.

Figure 1 is a schematic illustration of the apparatus according to a preferred exemplary embodiment of the present invention having a bent lamp and a cone reflector;

Figure 2 is a schematic illustration of the apparatus according to another preferred exemplary embodiment of the present invention having a linear flash-lamp and a straight reflector;

Figure 3 is a typical normalized output radiation spectrum in treatment mode with heating;

Figure 4 is a typical normalized output radiation spectrum in treatment mode with long wavelength cut off; and

Figure 5 is a typical normalized output radiation spectrum in diagnostic mode.

Before explaining at least one embodiment of the invention in detail, it is to be understood that the invention is not limited in its application to the details of construction and the arrangement of the components set forth in the following description or illustrated in the drawings. The invention is capable of other embodiments or being practiced or carried out in various ways. Also, it is to be understood that the phraseology and terminology employed herein is for the purpose of description and should not be regarded as limiting.

Generally, in the present invention, tumor necrosis consists of a composite effect of three processes. The first process is photodynamic therapy whose mechanism is initiated with the absorption of light by a photosensitive agent which has a much higher accumulation effect in cancer cells than in normal cells. Due to a photochemical reaction of incident light in the range of 600 to 1000 nm with this photosensitive agent, oxygen is generated in the singlet state. The reaction of this photochemically generated singlet oxygen with intracellular lipids, proteins, and nucleotides is deleterious to the cells and ultimately results in tumor necrosis.

The second process is pulsed heating of the tissue. The photochemical reaction of incident light with a photosensitive agent can be accelerated by heating tissue to a temperature which is significantly higher than normal temperature but lower than the temperature at which proteins coagulate. The absorption coefficient of skin or other tissue significantly depends upon the type of skin. Consequently, the spectrum of incident radiation and the power of light pulses is preferably varied over a wide range. The absorption coefficient of tissue is a function of light wavelength. Thus, for light in the radiation range of 600 to 700 nm, the depth of penetration into the dermis ranges up to approximately 1 mm. As a result, this range of radiation can be used to heat only a shallow surface layer of tissue or skin. Radiation having a wavelength range of 700 to 1200 nm can penetrate the dermis more deeply. Thus, tissue as deep as approximately 3 mm may be heated using such radiation. Because the epidermis absorption coefficient is significantly higher than that of the dermis, however, care should be taken to avoid overheating the epidermis. Transparent gel applied to the surface of the skin can prevent overheating.

The cooling time (t) of an object that has typical dimension (d) and diffusivity (a) can be written as:

$$t = d^2/a$$

Typically, the epidermis has cross dimensions of less than 0.1 mm. The diffusivity (a) is approximately $3 \times 10^{-7} \text{ m}^2 \text{ sec}^{-1}$. Thus, when gel is applied to the skin, the typical

cooling time of the epidermis will be approximately 33 msec. Gel application allows the epidermis to cool during the pulse delay and thus avoids adverse heating effects.

The third process used for tumor necrosis is to limit the blood supply to the tumor by coagulating the blood which results in lesion of the vessels surrounding the tumor. The absorption coefficient of blood is much higher than that of dermis or tissue for radiation in the wavelength range of 600 to 1000 nm. Hence at optimal chosen parameters of incident radiation, blood coagulation is possible without damaging the dermis and epidermis.

The composite effect of these three processes results in a more efficient necrosis of cancer cells. This combination decreases treatment time to approximately 5 minutes while simultaneously increasing the cancer necrosis effect.

Referring now to Figure 1, an operating head 100 encases a bent flashlamp 102, a silver coated conical reflector 104, and interchangeable interference filters 106. Filters 106 cut off the radiation spectrum at 520 nm. Transmission of radiation through filter 106 is dependent upon the incident angle of the radiation. Filter 106 reflects (does not transmit) any nonuseful incident radiation, thus avoiding overheating an absorbing filter 108.

Absorbing filter 108 and light guide 110 are external to operating head 100. Filter 108 cuts off radiation at 600 nm. Transmission through filter 108 of radiation having a wavelength less than 580 nm is less than 10^{-5} . Light guide 110 may be either a flexible light guide or a quartz light guide.

Flashlamp 102 may be operated in either CW mode or pulse repetitive mode. Reflector 104 focuses the light beam produced by flashlamp 102 and conducts the beam through interference filter 106 and absorbing filter 108 to light guide 110. Light guide 110 guides the light beam to treatment areas that are difficult to access, small targets, and internal tumors. Alternatively, light guide 110 may be decoupled from operating head 100. In such a configuration, the light beam produced by operating head 100 can be directly used to treat large, external tumors.

Referring now to Figure 2, an alternative exemplary embodiment of the invention is shown. Rather than using a bent flashlamp and conical reflector, this alternative embodiment employs a linear flashlamp 112 with a straight reflector 114.

When operating in the pulse mode, the invention produces a train of pulses at a repetition rate that varies from 0.1 to 10 pulses/second. The total number of pulses per pulse train can be selected in the range of 1 to 1000. The total dose to the treated area is the product of the number of pulses and the fluency per pulse.

In the therapeutic mode, two different spectral distributions can be selected. Referring now to Figure 3, the spectral distribution peaks at 615 nm and has a tail that reaches up to 1000 nm in therapeutic mode I. In addition, the spectral distribution of the radiation can be

controlled by varying the pulse parameters. For example, if it is necessary to increase the heating effect at a deeper depth, the long wavelength part of the radiation should be increased by decreasing the pulse power.

Referring now to Figure 4, in therapeutic mode II the spectrum of the radiation is cut off at 700 nm by an interference filter which is installed in place of absorbing filter 108. Due to this filter, no significant radiation is emitted having a wavelength greater than approximately 700 nm. Mode II can be used if minimum heating of tissue is desired.

The fluency generated in the therapeutic mode is a function of the distance between the face of light guide 110 and the treatment area. The operator can input this distance and the device calculates the fluency per pulse and the total dose corresponding to the selected distance.

The pulse duration can be varied in the range of 0.1 to 100 msec and the energy per pulse is variable in the range of 0.1 to 10 J/cm² (the 10 J/cm² is generated on the face of the light guide).

In the CW mode, the output radiation power density can be varied up to 1000 mW/cm². Heating effects are not a concern in this operating mode because the heating of the tumor and surrounding blood vessels is compensated by cooling due to the heat conductivity process.

For diagnostic purposes, both the CW mode and multiple pulse trains with a repetition rate of 0.01 to 0.2 pulse/sec may be used. The invention provides the necessary total light energy per pulse with radiation in the range of 350 to 500 nm with a variable pulse duration of 1 to 10 ms. The illuminated area may be as large as desired.

Figure 5 represents the normalized spectrum of radiation. As shown, the maximum level of radiation is at 400 nm. The necessary energy and power density may be obtained by varying the pulse repetition rate and energy per pulse. The exposed area can be modified by varying the distance from light guide 110 to the target. In the CW mode, the fluency rate can be as large as dozens of watts per square centimeter. For treatment of large areas, the light beam can be delivered directly to the target without light guide 110.

Operating parameters have been given above and are restated below, with alternatives. The operating parameters are exemplary and are not intended to be limiting.

In the diagnostic mode, the invention provides an output radiation including the following parameters:

- (1) either single pulse or repetitive pulse modes of operation with a frequency of 0.02 to 2 pulses/sec;
- (2) pulse duration that may be varied from 0.1 to 100 msec;
- (3) a spectrum of radiation in the range of 350 to 500 nm with a peak at 400 nm;
- (4) delivery of the light beam to the target area by

either a quartz or a flexible light guide;
 (5) optical energy density per pulse ranging from 0.02 to 4 J/cm² (in each pulse); and
 (6) an illumination area that varies in size dependent upon the distance of the target area from the light guide.

In the diagnostic mode, the invention can also be operated in CW mode, in which the suspected cancerous area is continuously illuminated while the physician looks for fluorescence that indicates the presence of a tumor. Illumination can be carried out through a light guide or by directly exposing the area to the opening in the housing that contains the lamp. The light guide is particularly useful for illuminating internal objects and areas that are difficult to access. When treating large surface areas, the light source may be used without the light guide.

In the PDT treatment mode, the invention provides the following output parameters:

- (1) continuous operation mode or repetitive mode with a frequency ranging from 0.1 to 1 pulse/sec;
- (2) pulse duration that may be varied from 0.1 to 100 msec;
- (3) spectrum of radiation in the range of either 600 to 1000 nm or 600 to 700 nm;
- (4) delivery of the light beam to the target either directly or by a flexible light guide if internal treatment is desired;
- (5) for external treatment of a large surface, the light beam can be delivered directly to the target without a light guide;
- (6) optical energy density per pulse ranging from 0.1 to 20 J/cm² (pulse) in repetitive mode;
- (7) fluency rate ranging from 100 to 2000 mW/cm²; and
- (8) illumination area that varies from 0.5 to 3 cm².

Thus, it should be apparent that there has been provided in accordance with the present invention a method and apparatus for the diagnostic and composite pulsed heating and photodynamic therapy treatment that fully satisfy the objectives and advantages set forth above. Although the invention has been described in conjunction with specific embodiments thereof, it is evident that many alternatives, modifications, and variations will be apparent to those skilled in the art. Accordingly, it is intended to embrace all such alternatives, modifications, and variations that fall within the spirit and broad scope of the appended claims.

Claims

1. An apparatus for therapeutically treating cancerous cells within a tissue having a plurality of vessels for supplying blood thereto, comprising:

a housing, including a window through which light can pass;
 a source of light, capable of being operated in both a continuous mode and a pulsed mode, disposed in the housing; and
 a reflector, disposed in the housing, that directs the light through the window.

2. An apparatus for therapeutically treating cancerous cells within a tissue having a plurality of vessels for supplying blood thereto, comprising:

a housing, including a window through which light can pass;
 a pulsed noncoherent source of light disposed in the housing; and
 a reflector, disposed in the housing, that directs the light through the window.

3. An apparatus for the therapeutic treatment and diagnosis of cancerous cells within a tissue comprising:

a light source;
 a filter system disposed in a path of the light to limit the light to a range of wavelengths; and
 a reflector, to focus the light beam and to conduct the light beam through the filter system.

4. The apparatus as claimed in claim 3 further including a light guide disposed to guide the filtered light to the tissue.

5. The apparatus as claimed in claim 2 wherein the light source may be operated in a continuous or pulsed mode.

6. The apparatus as claimed in claim 1 or 2 further including a filter disposed in a path of the light to provide a filtered light beam having a range of wavelengths.

7. The apparatus as claimed in claim 1 or 2 further including a light guide having a proximal end disposed near the window and a distal end capable of being disposed near the tissue.

8. The apparatus as claimed in claim 4 or 7 wherein the light guide is a flexible light guide.

9. The apparatus as claimed in claim 4 or 7 wherein the light guide is a quartz light guide.

10. The apparatus as claimed in claim 1 or 9 wherein the light source is a noncoherent source.

11. The apparatus as claimed in claim 10 wherein the light source is a source of pulsed light.

12. The apparatus as claimed in claim 10 or 11 wherein the light source when in the pulsed mode is capable of providing a selectable pulse frequency, a selectable pulse duration, and a selectable energy density per pulse, and further wherein the number of pulses may be varied. 5
13. The apparatus as claimed in claim 3 or 10 wherein the light source is a source of continuous wave having a selectable power density. 10
14. The apparatus as claimed in claim 3 or 10 wherein the filter system includes a first interference filter and an absorbing filter. 15
15. The apparatus as claimed in claim 1 or 2 wherein the light source is a linear flashlamp and the reflector is a straight reflector.
16. The apparatus as claimed in claim 1 or 2 wherein the light source is a bent flashlamp and the reflector is a conical reflector. 20

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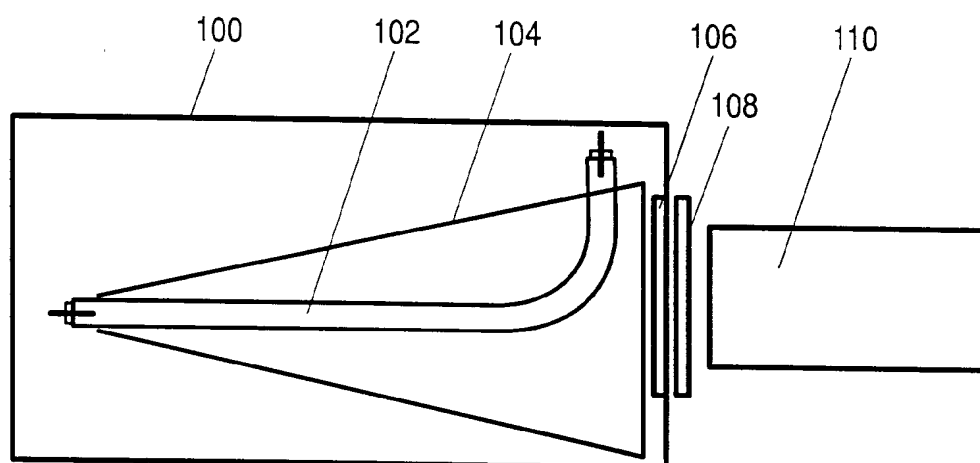


Fig. 1

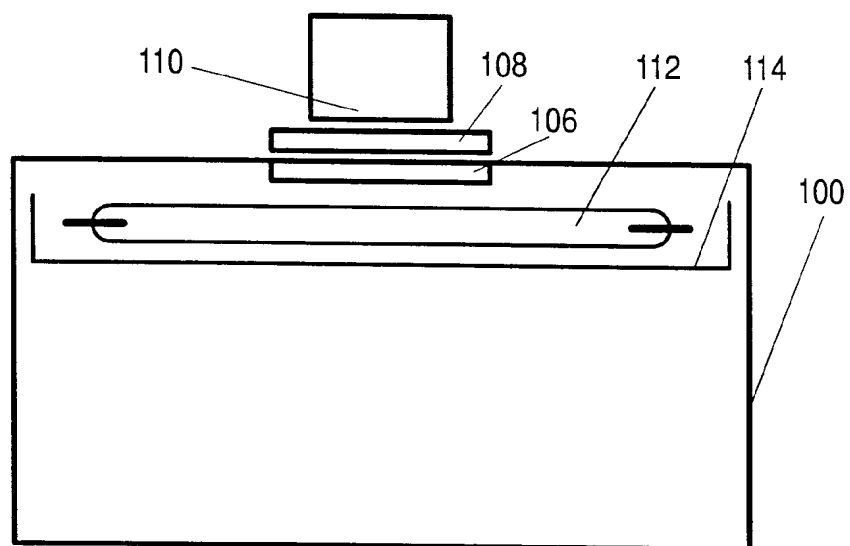


Fig. 2

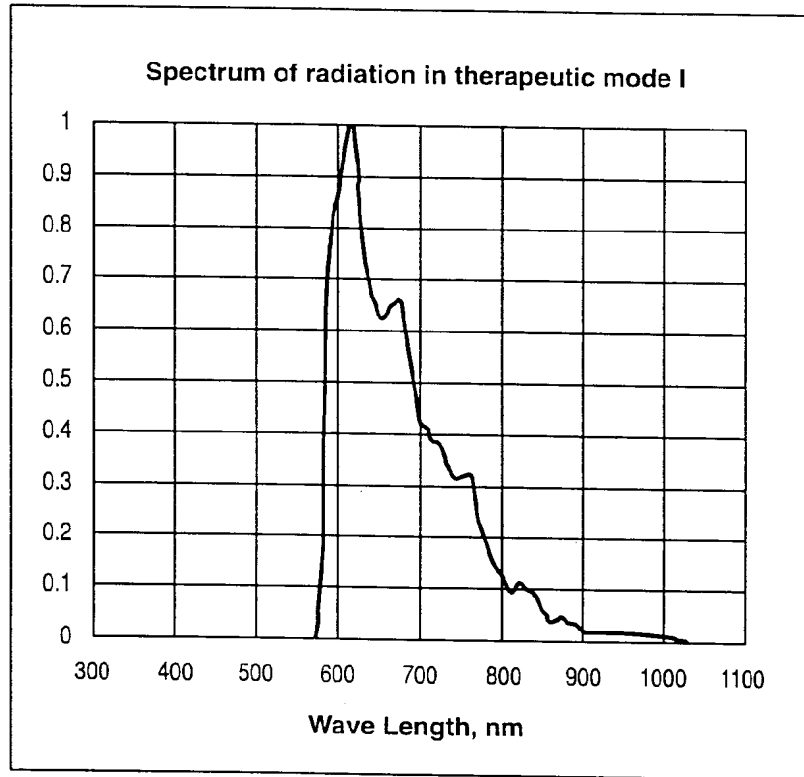


Fig. 3

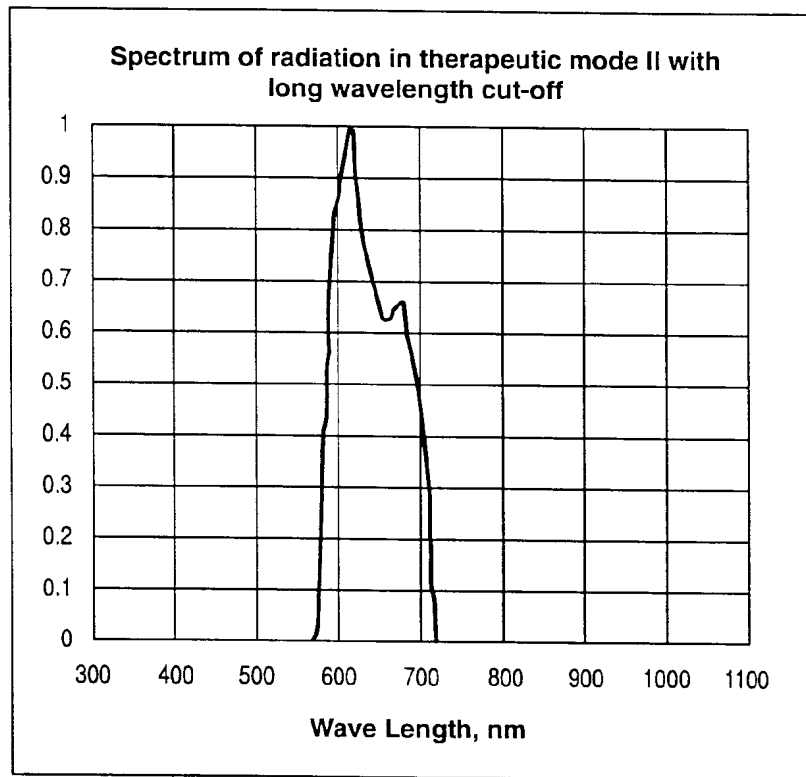


Fig. 4

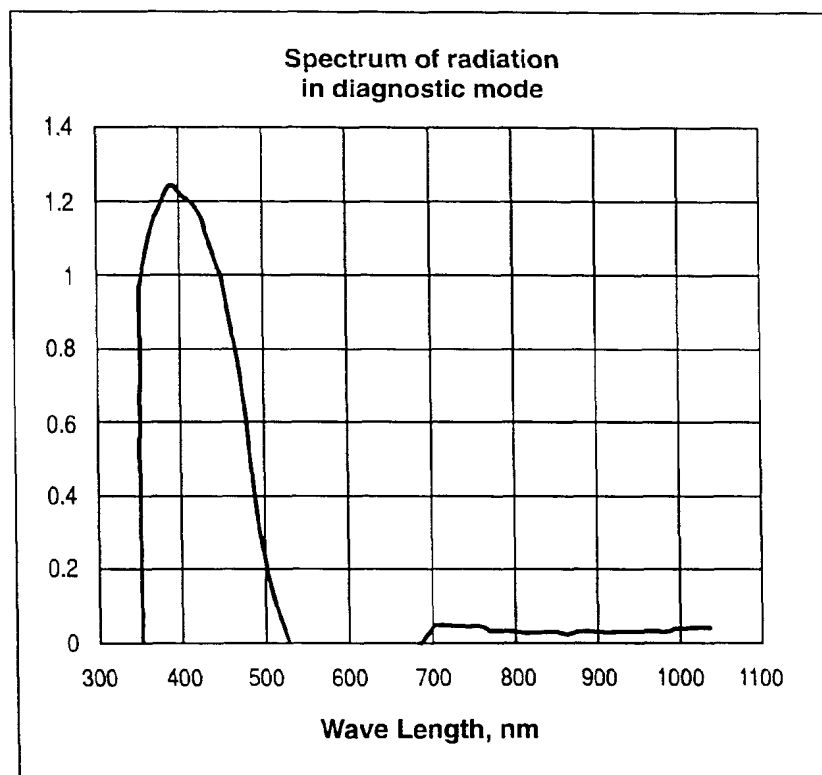


Fig. 5



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(54) **Method and apparatus for depilation using pulsed electromagnetic radiation**

(57) A method and apparatus for removing hair is disclosed. The method includes the step of producing at least one pulse of incoherent electromagnetic energy. The incoherent electromagnetic energy is then coupled to an area of surface of the tissue that includes more than one hair follicle. The method may alternatively include the step of applying a gel (103) on a surface of the tissue to cool the tissue. The energy heats the hairs and

hair follicles, without heating the tissue. The apparatus includes a source (301) of pulsed incoherent electromagnetic energy. The source is located within a housing (302), and a coupler (310) directs the incoherent electromagnetic energy to the surface of the tissue. In an alternative arrangement a gel disposed on the surface of the tissue such that the gel cools the tissue but is not adjacent, and does not cool, the hair follicle.

Description

The invention relates generally to devices and methods for removing hair, and more particularly to such devices and methods that utilize electromagnetic energy to kill hair follicles.

Hair can be removed permanently for cosmetic reasons by various methods, for example by heating the hair and the hair follicle to a high enough temperature that results in their coagulation. It is known that blood is coagulated when heated to temperatures of the order of 70°C. Similarly, heating of the epidermis, the hair and the hair follicle to temperatures of the same order of magnitude will also cause their coagulation and will result in permanent removal of the hair.

One common method of hair removal, often called electrolysis, is based on the use of "electric needles" that are applied to each individual hair. An electrical current is applied to each hair through the needle. The current heats the hair, causes its carbonization and also causes coagulation of the tissue next to the hair and some coagulation of the micro-vessels that feed the hair follicle.

While the electrical needle method can remove hair permanently or long term, its use is practically limited because the treatment is painful and the procedure is generally tedious and lengthy.

Light can also be used effectively to remove hair. For example, other prior art methods of hair removal involve the application of pulsed light, generally from coherent sources such as lasers. R. A. Harte, et al., in U.S. Patent No. 3,693,623, and C. Block, in U.S. Patent No. 3,834,391, teach to remove hair by coagulating single hair with a light coupled to the individual hair by an optical fiber at the immediate vicinity of the hair. Similarly, R. G. Meyer, in U.S. Patent No. 3,538,919, removes hair on a hair by hair basis using energy from a pulsed laser. Similar inventions using small fibers are described in U.S. Patent No. 4,388,924 to H. Weissman, et al. and U.S. Patent No. 4,617,926 to A. Sutton. Each of these teach to remove hair one hair at a time, and are thus slow and tedious.

U.S. Patent No. 5,226,907, to N. Tankovich, describes a hair removal method based on the use of a material that coats the hair and hair follicle. The coating material enhances absorption of energy by the follicles, either by matching the frequency of a light source to the absorption frequency of the material, or by photochemical reaction. In either case the light source is a laser. One deficiency of such a method and apparatus is that lasers can be expensive and subject to stringent regulations. Additionally, the coating material must be applied only to the hair follicles, to insure proper hair removal and to prevent damage of other tissue.

Light (electromagnetic) energy used to remove hair must have a fluence such that sufficient energy will be absorbed by the hair and the hair follicle to raise the temperature to the desired value. However, if the light is ap-

plied to the surface of the skin other than at the precise location of a hair follicle, the light will also heat the skin to coagulation temperature and induce a burn in the skin.

Accordingly, it is desirable to be able to effectively heat multiple follicles, without burning the surrounding skin. Such a method and apparatus should be able to remove more than one hair at a time, and preferably over a wide area of skin, for example at least two square centimeters. Additionally, the method and apparatus should be capable of using incoherent light.

In accordance with one aspect of the invention, a method of removing hair from an area of tissue includes producing at least one pulse of incoherent electromagnetic energy. The incoherent electromagnetic energy is then coupled to an area of the surface of the tissue that includes more than one hair follicle.

Additionally, in one alternative embodiment the energy may, but not necessarily, be produced by pulsing a flashlamp to generate a pulse having an energy fluence on the order of 10 to 100J/cm². The energy can be coupled through a window in a housing in which the flashlamp is located, by reflecting the energy to the tissue through the window and through a gel located on a surface of the tissue. The window may be brought into contact with the gel. In other alternative embodiments the angular divergence of the electromagnetic energy is controlled, and thus the depth of penetration into the tissue, and the coupling to the hair and to the hair follicles, is also controlled. In another alternative embodiment each step of the method is repeated, but at least two angular divergences are used, thus obtaining at least two depths of penetration.

In other alternative embodiments electromagnetic energy is filtered. Specifically, in one embodiment the electromagnetic energy is filtered according to the pigmentation level of the tissue to be treated. In another alternative energy that has a wavelength of less than 550nm and greater than 1300nm is filtered. Some or all of such energy can be filtered.

In yet another alternative embodiment the pulse produced has a width of less than 200msec, and/or the delay between pulses is of the order of 10 to 100msec between the pulses. In one embodiment the surface area of the energy at the tissue is at least two square centimeters.

In accordance with a second aspect of the invention an apparatus for removing hair from an area of tissue that includes more than one hair follicle includes a source of pulsed incoherent electromagnetic energy. The source is located within a housing, and a coupler directs the incoherent electromagnetic energy to the surface of the tissue.

According to an alternative embodiment the source is a flashlamp and a pulse generating circuit that generates pulses of energy that have an energy fluence on the order of 10 to 100J/cm². The coupler can include a transparent window and the housing a reflective interior,

wherein the energy is reflected to the window. A gel is disposed on the surface of the tissue and the window is in contact with the gel, to couple the energy through the window and gel to the surface of the tissue. In another alternative embodiment the energy provided by the coupler has a range of angular divergences.

In another alternative embodiment at least one band-pass electromagnetic filter is disposed between the source and the tissue. The filter can be selected such that the wavelength of the energy that passes through the filter is based on the pigmentation level of the treated tissue. Alternatively, the filters pass energy that has a wavelength of between 550nm and 1300nm.

In other embodiments source provides pulses having a width of less than 200msec, and/or delays between pulses on the order of 10 to 100msec. In another embodiment the area of the energy at the tissue is at least two square centimeters.

According to a third aspect of the invention, a method of removing hair from an area of tissue that has more than one hair follicle includes producing at least one pulse of electromagnetic energy. A gel on a surface of the tissue cools the tissue, but the gel is not adjacent the hair follicle. The electromagnetic energy is coupled to the surface of the tissue.

In one alternative embodiment the energy is produced by pulsing a flashlamp, and a pulse having an energy fluence on the order of 10 to 100J/cm² is thereby generated. In another embodiment the flashlamp is located in a housing that includes a transparent window and the energy is reflected through the window and directed through the gel to the tissue. In yet another alternative embodiment the angular divergence of the electromagnetic energy is selected to determine the depth of penetration into the tissue, and to determine the coupling to the hair and to the hair follicles. Also, each step of the method may be repeated using at least two different angular divergences, whereby at least two depths of penetration are obtained.

In another alternative embodiment the electromagnetic energy is filtered. The filtering can be done in accordance with the pigmentation level of the treated tissue. Alternatively, filtering may include filtering some or all of the energy that has a wavelength of less than 550nm and greater than 1300nm.

In another alternative embodiment pulses produced have a width of less than 200msec. The delay between pulses may be on the order of 10 to 100msec. Also, the area of the energy at the tissue can be large, for example more than two square centimeters. The energy may be incoherent, such as that produced by a flashlamp for example, or coherent, such as that produced by a laser, for example.

In accordance with a fourth aspect of the invention an apparatus for removing hair from an area of tissue that has more than one hair includes a source of pulsed electromagnetic energy. A gel is disposed on the surface of the tissue such that the gel cools the tissue but

is not adjacent, and does not cool, the hair follicle. A coupler is disposed between the source and the surface to couple the energy to the surface.

In one alternative embodiment the source is a pulsed flashlamp that generates pulses having an energy fluence on the order of 10 to 100J/cm². In another alternative the flashlamp is located in a housing that includes a transparent window and a reflective interior. In yet another alternative embodiment the shape of the coupler determines the angular divergence of the electromagnetic energy, which determines the depth of penetration of the energy into the tissue, and determines the coupling to the hair and to the hair follicles. The apparatus may include a band-pass filter disposed between the source and the surface. In one alternative the band-pass filter passes energy having a wavelength of between 550nm and 1300nm. The source may be a source of incoherent energy, or a source of coherent energy, such as a laser, for example.

The present invention will now be described, by way of example only, with reference to the accompanying drawings, in which like numerals designate corresponding elements or sections throughout, and in which:

Figure 1 is a schematic drawing of a cross section of a hair follicle in the dermis and a gel applied to the epidermis in accordance with the present invention;

Figure 2 is a graph showing the optical properties of the skin;

Figure 3 is a side view of a hair removal apparatus constructed in accordance with the present invention;

Figure 4 is a front view of a hair removal apparatus constructed in accordance with the present invention;

Figure 5 is a divergent coupler such as one used in the present invention; and

Figure 6 is a non-divergent coupler such as one used in the present invention.

Before explaining at least one embodiment of the invention in detail it is to be understood that the invention is not limited in its application to the details of construction and the arrangement of the components set forth in the following description or illustrated in the drawings. The invention is capable of other embodiments or of being practiced or carried out in various ways. Also, it is to be understood that the phraseology and terminology employed herein is for the purpose of description and should not be regarded as limiting.

Generally, in the present invention, hair is removed by exposing the "hairy" area to intense, wide area, pulsed electromagnetic (light) energy. The energy heats the hair and coagulates the tissue around the hair and follicle without damaging the healthy skin.

An optically transparent water based gel may be applied to the skin prior to treatment. As used herein gel

means a viscous fluid that is preferably, but not necessarily water based. The gel is used to cool the epidermis which is the primary location of light absorption by tissue, due to the melanin content of the epidermis. The gel is applied so as not to penetrate into the cavity generated by the hair follicle, and thus does not cool the hair and the hair follicle. As a result the energy is selectively applied to coagulate the hair without damaging the skin.

A polychromatic light source, such as a high intensity pulsed flashlamp, is an example of a source suitable for the purposes described herein. One advantage of a polychromatic source such as a flashlamp is that energy having a wavelength in the range of 550 to 630nm is heavily absorbed in blood and can be used to coagulate the vessel that feeds the hair. Additionally, longer wavelengths, in the range of 600 to 1100nm have a very good penetration into non-pigmented skin. This wavelength range can be used to couple to the melanin of the hair. The higher pigmentation of the hair and the hair follicle can enhance the absorption of energy by the hair.

Flashlamps also have the advantage of being able to illuminate a large area, thus minimizing the treatment time. The flashlamp combined with a proper reflector can deliver the required fluences to areas on the order of a few square centimeters in a single application. However, other light sources, such as pulsed lasers can be used as well.

Referring now to Figure 1, a schematic drawing of a cross section of a hair follicle 100 in a dermis 102 is shown. As may be seen in Figure 1, a gel 103 applied to an epidermis 104. In the present invention, water based transparent gel 103 is applied to a large section of the skin that is covered by hair, such as hair 105. Gel 105 is applied to epidermis 104 and creates a thin layer on top of epidermis 104. This layer is closely coupled to epidermis 104 and acts as a heat sink that cools epidermis 104 when light (electromagnetic energy) is applied to the area. As may also be seen in Figure 1, gel 103 does not penetrate into a cavity 106 formed by hair follicle 100 due to its surface tension properties and the fact that the hair is naturally covered by a thin layer of fatty material which makes it hydrophobic. The much higher heat diffusivity of gel 103 compared to that of air which fills cavity 106 enables fast cooling of epidermis 104, represented by arrows 107, while hair 105 is cooled at a much slower rate.

The cooling time - δt of an object that has typical dimensions d and diffusivity - α can be written as:

$$\delta t \approx d^2 / 16\alpha$$

The epidermis has typical cross dimensions of less than 0.1mm, which is also the typical diameter of hair. The diffusivity of water is approximately $\alpha = 3 \times 10^{-9} \text{m}^2 \text{sec}^{-1}$.

The gel is applied, in the manner shown in Figure 1, over a wide area. When the gel is so applied the typical cooling time of the hair will be on the order of

200msec and that of the epidermis will be on the order of 5msec. This difference in cooling times is due to the fact that the gel does not penetrate into the hair follicles. It is preferable to use a transparent gel since the gel acts only as a cooling agent and should not be heated by the external illumination.

In accordance with the invention, light is applied to the treated area in either a long pulse or in a sequence of pulses separated by a delay. The delay and/or pulse length is preferably controlled by the operator to provide enough heat to remove the hair but not enough heat to damage the skin. For example, the pulse length or delay between the pulses should be more than the cooling time of the gel covered epidermis and less than the cooling time of the hair and follicle. Thus, referring to the above discussion on cooling times, a pulse length of 50msec if a single pulse is used or a delay of 50msec between the pulses if a pulse sequence is used are appropriate values.

The spectrum of the light source may be selected with reference to the absorption by the skin, by the hair and by the blood vessels feeding the hair. For example, the hair follicle has typical a depth of 1 to 2mm. It is preferable, therefore, to use a light wavelength range that can penetrate into this depth without very high attenuation.

Figure 2 is a graph showing the scattering, absorption and effective attenuation coefficients in fair skin dermis and the absorption coefficient of blood in the 400 to 1000nm range. Because a wide area is illuminated, rather than a single hair, it is preferable to use a wavelength range that penetrates into the skin without being highly attenuated. The skin attenuation coefficient controls the depth of penetration of light into the skin. As may be seen in Figure 2 wavelengths that are longer than 550nm will be more effective to penetrate deep enough into the skin. Shorter wavelengths are less desirable because they will be highly attenuated before reaching the lower parts of the hair follicles.

Wavelengths significantly longer than 1,000nm are also less effective due to high absorption of infrared in water which constitutes more than 70% of skin. Wide area photo thermal hair removal of the present invention preferably uses light that can penetrate deep into the skin, since light is coupled to the hair and the hair follicles only after it penetrates through the skin. Most of the spectrum of light at wavelengths longer than 1,300nm is heavily absorbed in water and will be less useful because it does not penetrate very deep into the skin. For example, CO₂ laser radiation in the 10,000nm range penetrates only a few tens of microns into the skin.

Referring now to Figures 3 and 4, one preferred embodiment of hair remover 300 includes a flashlamp 301 located in a housing 302 having a handle. The flashlamp is shown adjacent gel 103 and hairy skin 102/104/105. One flashlamp that the inventors have found effective for hair removal is described in detail in co-pending United States Patent Application For Method and Apparatus

For Therapeutic Electromagnetic Treatment, Serial No. 07/964,210, filed October 20, 1992, and incorporated herein by reference. The flashlamp described therein provides a suitable fluence and it illuminates a large area in a single pulse (on the order of 10X50 mm).

Such a flashlamp is driven by a variable pulse width power source. The flashlamp is contained in housing 302 and the light from the flashlamp is directed towards the skin by a reflector 305 that has a high reflectivity.

Also shown in Figures 3 and 4 is a filter 307, that is disposed between flashlamp 301 and gel 103. The filter, or in an alternative embodiment, multiple filters, are used to control the spectrum generated by the light source. As used herein filter, or band-pass filter, describes a device that allows electromagnetic energy (light) of certain wavelengths or frequencies to pass. The other wavelengths or frequencies are either partially or wholly removed.

The operator can select the filter according to the skin pigmentation of the person being treated. For the embodiment using a flashlamp, one can take advantage of the spectral range typically generated by such a lamp, which is in the range of 200 to 1300nm for high pressure xenon flashlamps operated at high current densities (on the order of 1,000 to 5,000 A/cm²). Since hair removal is mainly done for cosmetic reasons and is mostly important for cases of darker hair, the hair itself will absorb light in a wide spectral range in the visible and the near infrared. The shorter wavelengths generated by the flashlamp may be removed since they do not penetrate as deeply into the skin (as can be seen from Figure 2).

In one embodiment a long pass filter that transmits only wavelengths longer than the cut off wavelength of the filter is used. A cut off wavelength of 600nm is used in a preferred embodiment when the person being treated has fair skin. A cut off wavelength in the range of 700 to 800nm is used in the preferred embodiment to treat people with dark skin. According to the invention, the filters may be, for example, dichroic filters or absorbing filters. The desired spectrum can also be achieved by more than one filter or by band-pass filters.

Light from flashlamp 301 is coupled to the skin through a transparent window 308 and a coupler 310 (described below). As shown in Figures 3 and 4, window 308 is placed on transparent water based gel 103. In use, the operator holds hair remover 300 by handle 304, and places it on the area of skin where treatment is desired (and gel 103 has been applied). Transparent window 308 creates a well defined flat surface on gel 103, through which light enters into gel 103 and into the skin.

The operator selects the pulse and energy fluence parameters on a control unit (not shown). The power and control unit are preferably housed in a separate box and will include power from a capacitor charged to a high voltage by a DC power supply, wherein the capacitor is discharged through the flashlamp. Hair remover 300 can be connected to the power and control unit via a flexible cable that allows easy aiming of the device when

aiming it to the treatment area on the patient's skin. Pulse length control can be achieved by using a few pulse forming networks that can generate different pulse widths. Alternatively, an opening 309 may include a solid state opening switch that can stop the discharge at a time preset by the operator, thus controlling the pulse width. These elements of the device are well known and can be easily constructed, or replaced by similar elements, as one skilled in the art will know.

After the parameters have been selected, the operator fires the unit by pressing a switch that can be located in a variety of locations.

A total fluence on the order of 10 to 100J/cm² will successfully remove the hair. This fluence can be determined from the requirement of reaching a high enough temperature of the hair and hair follicle, and considering the penetration of light, through the skin and into the hair and hair follicle, absorption of light in the hair and hair follicle, specific heat capacity of the hair and the hair follicle, and the cooling of the hair during the pulse by heat conductivity to the surrounding skin.

Coupler 310 transmits light from flashlamp 301 to gel 103 and to the skin. The coupler can be comprised of a hollow box with internally reflecting walls that act as a light guide for the light generated by flashlamp 301, to transmit the light (electromagnetic energy) to the skin. Coupler 310 may alternatively be made from other material, for example, a solid transparent material such as glass or acrylic in which light reflection from the walls is achieved by using total internal reflection on the side walls.

Coupler 310 is used, in one alternative embodiment, to control the angular distribution of the light rays impinging on the skin. Light rays will hit the hair or the hair follicle predominantly when they are travelling in a direction perpendicular to the plane of the skin. A distribution of light rays that has a relatively wide angular divergence when treating shallow hair is desirable to direct a large portion of the energy to the hairs and follicles. Conversely, a narrow divergence is preferable when deep penetration is desired.

In one embodiment both shallow and deep penetration is obtained by using a two stage treatment process. A narrow divergence beam is used first to treat the deeper hair follicles, while a high divergence beam is used to treat the top of the hair follicles.

Figure 5 shows a coupler 501 having an exit beam with a greater angular divergence than that of the entrance beam. As shown in Figure 5, a beam 502 enters coupler 501 at a small angle, relative to the axis of coupler 501. When beam 502 exits coupler 501 the angle, relative to the axis, is much greater. The tapered shape of coupler 501 enhances this divergence.

Figure 6 shows a straight coupler 601, that maintains the angular distribution of the rays of light that enter into it. A beam 602 is shown entering and exiting coupler 601 with the same angle, relative to the axis of coupler 601. Alternate use of both couplers 501 and 601 can

achieve the narrow and deep penetration discussed above. Alternatively, the user can select the type of coupler according to the depth of hair being treated.

Clinical tests have been performed on hair on the legs of a few patients. Hair was removed for at least two months without observing any hair growing back on the exposed areas during this period. The experiments were performed with high fluences, i.e., up to 45J/cm² in each exposure. The spectrum used covered the range of 570 to 1100nm and the fluence was supplied in a triple pulse with delays of 50 to 100msec between pulses. The pulse sequence enabled hair removal with minimum pain and no damage to the skin. The transparent gel that was used in these experiments was a water based ultrasound gel, such as that commonly available.

Thus, it should be apparent that there has been provided in accordance with the present invention a flashlamp and coupler that fully satisfy the objectives and advantages set forth above. Although the invention has been described in conjunction with specific embodiments thereof, it is evident that many alternatives, modifications and variations will be apparent to those skilled in the art. Accordingly, it is intended to embrace all such alternatives, modifications and variations that fall within the spirit and broad scope of the appended claims.

Claims

1. A method of removing hair from an area of tissue that includes a plurality of hair follicles, characterised by the steps of:
 - producing at least one pulse of incoherent electromagnetic energy;
 - and coupling the incoherent electromagnetic energy to the surface of the tissue.
2. A method as claimed in claim 1, wherein the step of producing comprises the step of pulsing a flashlamp to generate a pulse having an energy fluence on the order of 10 to 100J/cm².
3. A method as claimed in claim 2, wherein the flashlamp is located in a housing that includes a transparent window and wherein the step of coupling includes the steps of reflecting the energy to the tissue, through the window and a gel located on a surface of the tissue.
4. A method as claimed in claim 3, wherein the step of coupling further includes bringing the window into contact with the gel.
5. A method as claimed in claim 4, wherein the step of coupling includes controlling the angular divergence of the electromagnetic energy, whereby the depth of penetration into the tissue, and the coupling to the hair and to the hair follicles, is controlled.
6. A method as claimed in claim 5, wherein each step of the method is repeated at least twice, and wherein at least two angular divergences are used, whereby at least two depths of penetration are obtained.
7. A method as claimed in any one of claims 1 to 6, wherein the step of producing comprises the step of filtering the electromagnetic energy.
8. A method as claimed in claim 7, wherein the step of filtering includes the step of filtering electromagnetic energy according to the pigmentation level of the treated tissue.
9. A method as claimed in claim 7, wherein the step of filtering includes the step of filtering energy that has a wavelength of less than 550nm and greater than 1300nm.
10. A method as claimed in claim 9, wherein the step of filtering energy that has a wavelength of less than 550nm and greater than 1300nm, includes filtering substantially all such energy.
11. A method as claimed in any one of the preceding claims, wherein the step of producing includes the step of producing at least one pulse having a width of less than 200msec.
12. A method as claimed in claim 11, in which the step of producing includes producing a plurality of pulses with a delay of the order of 10 to 100msec between the pulses.
13. A method as claimed in any one of the preceding claims, in which the area of the energy at the tissue is at least two square centimeters.
14. An apparatus for removing hair from an area of tissue including a plurality of hair follicles, characterised in that the apparatus comprises a source of pulsed incoherent electromagnetic energy, wherein the source is located within a housing (302) that includes a coupler capable of being disposed to direct the incoherent electromagnetic energy to the surface of the tissue.
15. An apparatus as claimed in claim 14, wherein the source includes a flashlamp (301) and a pulse generating circuit capable of generating a pulse having an energy fluence on the order of 10 to 100J/cm².
16. An apparatus as claimed in claim 14 or 15, wherein the housing includes a reflective interior and a transparent window, wherein the energy is reflected

through the coupler to the window, and further wherein a gel (103) is disposed on the surface of the tissue and the window is in contact with the gel.

17. An apparatus as claimed in claim 16, wherein the coupler is a tapered coupler. 5
18. An apparatus as claimed in any one of claims 14 to 17, further including a band-pass electromagnetic filter (307) disposed between the source and the tissue. 10
19. An apparatus as claimed in claim 18, further including a plurality of band-pass electromagnetic filters capable of being disposed between the source and the tissue, wherein a filter is selected such that the wavelength of the energy that passes through the filter is based on the pigmentation level of the treated tissue. 15
20. An apparatus as claimed in claim 18 wherein the filter passes energy that has a wavelength of between 550nm and 1300nm. 20
21. An apparatus as claimed in any one of claims 14 to 20, wherein the source is a source of pulses having a width of less than 200msec. 25
22. An apparatus as claimed in any one of claims 14 to 21, wherein the source is a source of pulses having a delay on the order of 10 to 100msec between the pulses. 30
23. An apparatus as claimed in any one of claims 14 to 22, in which the area of the energy at the tissue is at least two square centimeters. 35
24. A method of removing hair from an area of tissue including a plurality of hair follicles, characterised by the steps of: 40
producing at least one pulse of electromagnetic energy;
providing a gel on a surface of the tissue, wherein the gel is not adjacent the hair follicle, and further wherein the gel cools the tissue; and coupling the electromagnetic energy to the surface of the tissue. 45
25. A method as claimed in claim 24, wherein the step of producing comprises the step of pulsing a flash-lamp to generate a pulse having an energy fluence on the order of 10 to 100J/cm². 50
26. A method as claimed in claim 25, wherein the flash-lamp is located in a housing that includes a transparent window and wherein the step of coupling includes the steps of reflecting the energy to the tissue. 55

sue through the gel.

27. A method as claimed in claim 25, wherein the step of coupling includes controlling the angular divergence of the electromagnetic energy, whereby the depth of penetration into the tissue, and the coupling to the hair and to the hair follicles, is controlled.
28. A method as claimed in claim 27, wherein each step of the method is repeated at least twice, and wherein at least two angular divergences are used, whereby at least two depths of penetration are obtained.
29. A method as claimed in any one of claims 24 to 28, wherein the step of producing comprises the step of filtering the electromagnetic energy.
30. A method as claimed in claim 29, wherein the step of filtering includes the step of filtering electromagnetic energy according to the pigmentation level of the treated tissue.
31. A method as claimed in claim 29, wherein the step of filtering includes the step of filtering some of the energy that has a wavelength of less than 550nm and greater than 1300nm.
32. A method as claimed in any one of the preceding, wherein the step of producing includes the step of producing at least one pulse having a width of less than 200msec.
33. A method as claimed in claim 32, in which the step of producing includes producing a plurality of pulses with a delay on the order of 10 to 100msec between the pulses.
34. A method as claimed in any one of claims 24 to 33, in which the area of the energy at the tissue is at least two square centimeters.
35. A method as claimed in any one of claims 24 to 34, in which the step of producing includes the step of producing coherent energy.
36. An apparatus for removing hair from an area of tissue that includes a plurality of hair follicles, characterised in that the apparatus comprises:
a source of pulsed electromagnetic energy;
a gel (103) disposed on the surface of the tissue, wherein the gel is not adjacent the hair follicle, and further wherein the gel cools the tissue; and
a coupler (310) disposed between the source and the surface, wherein, the electromagnetic energy is coupled to the surface of the tissue.

37. An apparatus as claimed in claim 36, wherein the source is a pulsed flashlamp (301) capable of generating pulses having an energy fluence on the order of 10 to 100J/cm². 5
38. An apparatus as claimed in claim 37, wherein the flashlamp is located in a housing that includes a transparent window and wherein the housing includes a reflective interior. 10
39. An apparatus as claimed in claim 37 or 38, wherein the shape of the coupler determines the angular divergence of the electromagnetic energy, whereby the depth of penetration into the tissue, and the coupling to the hair and to the hair follicles, is controlled. 15
40. An apparatus as claimed in any one of claims 37 to 39, further including a band-pass filter disposed between the source and the surface. 20
41. An apparatus as claimed in claim 40, wherein the band-pass filter passes energy having a wavelength of between 550nm and 1300nm.
42. An apparatus as claimed in any one of claims 36 to 25

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FIG. 1

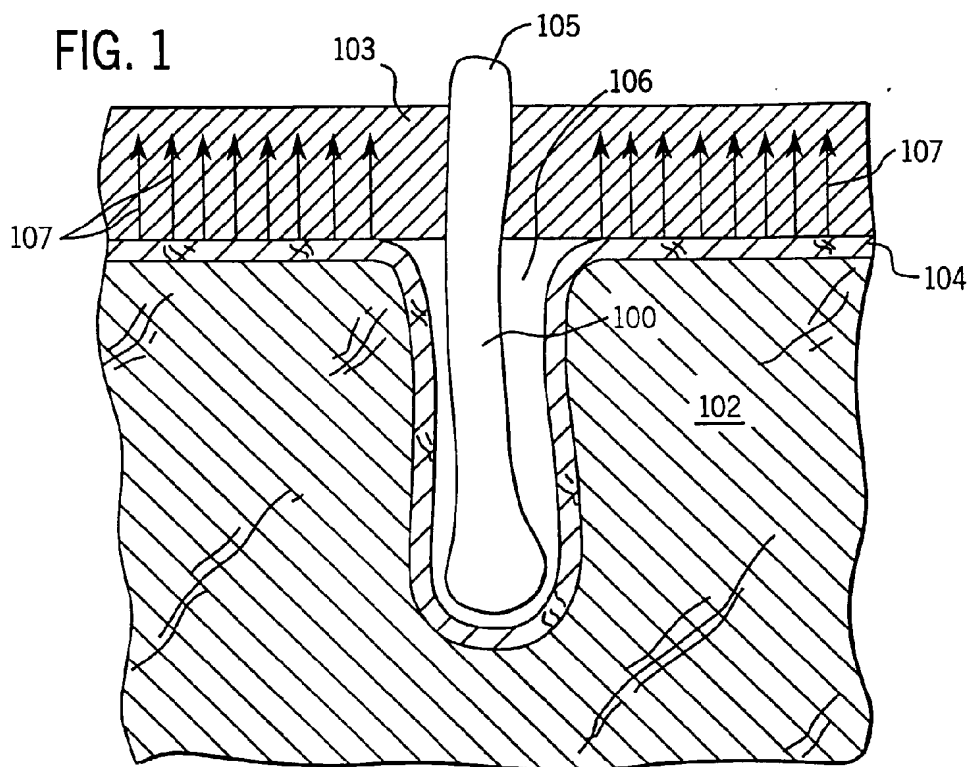


FIG. 5

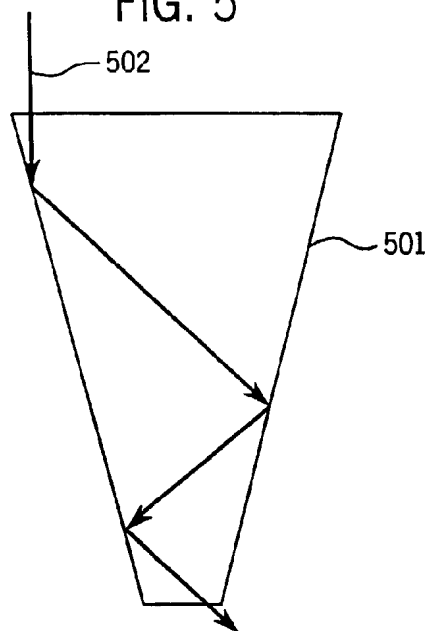


FIG. 6

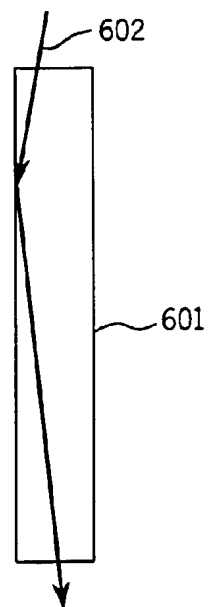
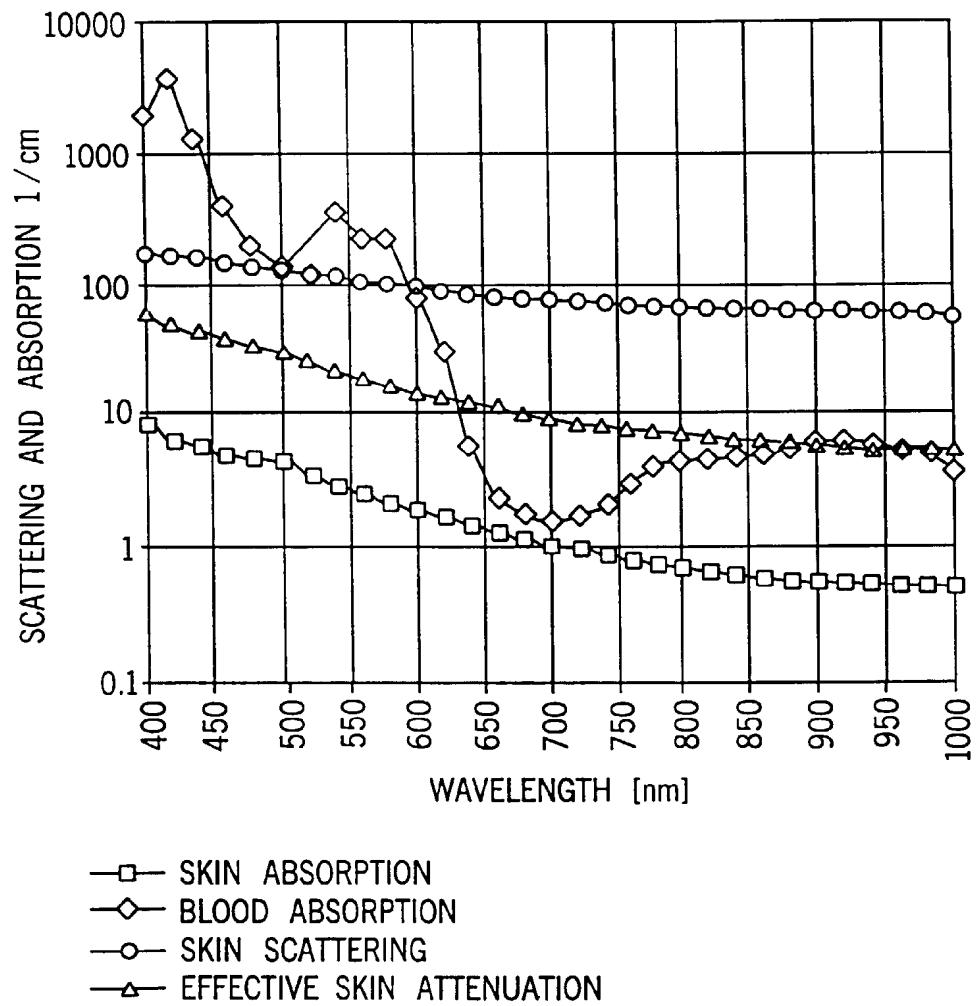
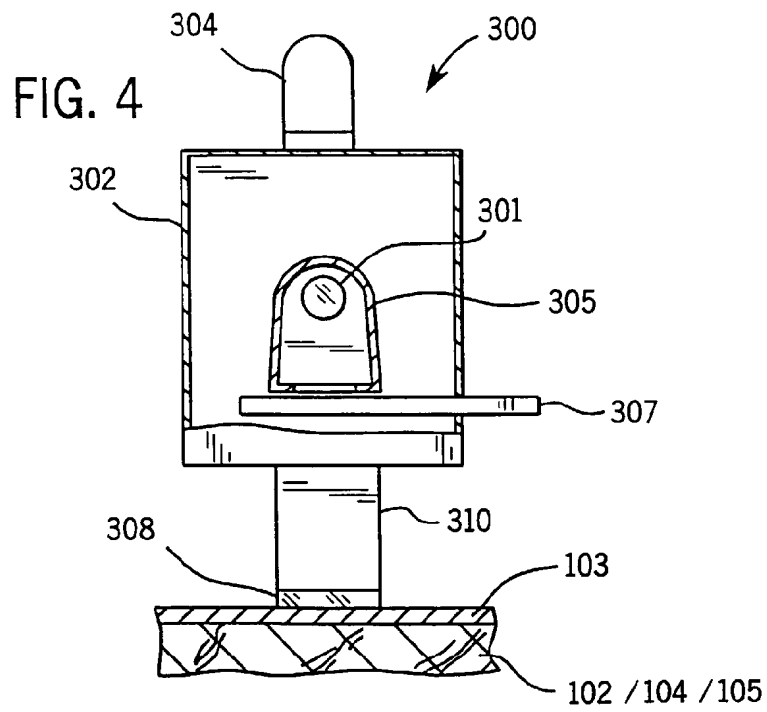
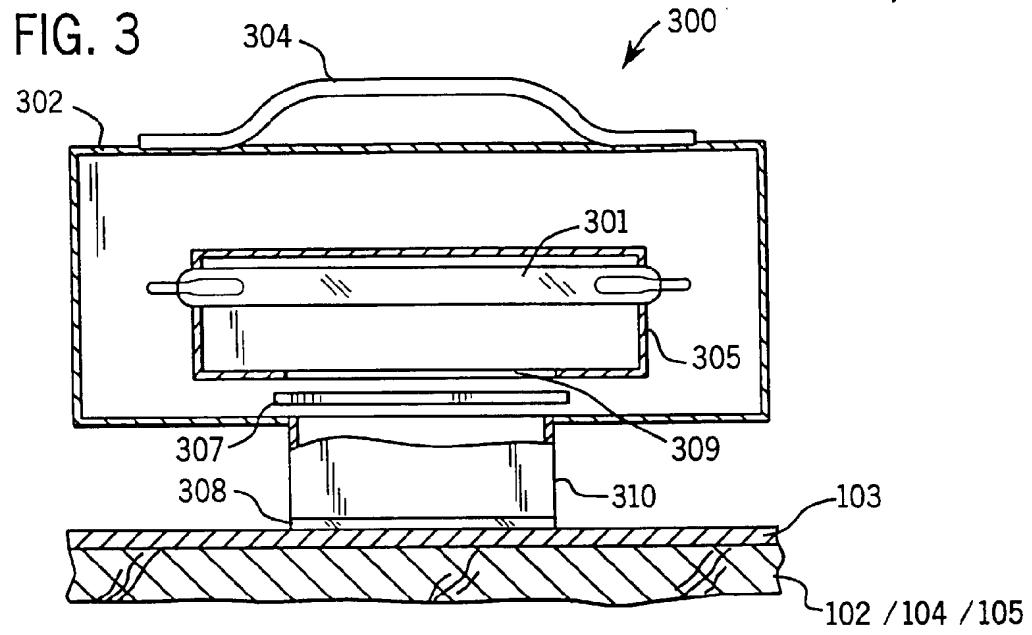


FIG. 2







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(54) **Method and apparatus for therapeutic electromagnetic treatment**

(57) A therapeutic treatment device is disclosed having a pulsed incoherent light source connected to a microprocessor. The microprocessor is connected to a

display and a user interface. It also has a means to receive input parameters related to the treatment, as well as a means to provide output parameters related to the treatment on the display.

Description

FIELD OF THE INVENTION

The present invention relates generally to the art of therapeutic electromagnetic treatment and more specifically to a method and apparatus for utilizing a spatially extended pulsed light source such as a flashlamp and providing treatment parameters for its use.

RELATED APPLICATIONS

This application is a continuation-in-part of prior co-pending United States application Serial No. 08/477,479, filed June 7, 1995, entitled "Method And Apparatus For Therapeutic Electromagnetic Treatment," and is a continuation of prior co-pending United States application Serial No. 08/473,532, filed June 7, 1995, entitled "Method And Apparatus For Therapeutic Electromagnetic Treatment," which are continuations of prior co-pending United States application Serial No. 08/383,509, filed February 3, 1995, entitled "Method And Apparatus For Therapeutic Electromagnetic Treatment," which is a continuation-in-part of prior co-pending United States application Serial No. 07/964,210, filed October 20, 1992, entitled "Method And Apparatus For Therapeutic Electromagnetic Treatment," issued on April 11, 1995, as U.S. Patent No. 5,405,368.

BACKGROUND OF THE INVENTION

It is known in the prior art to use electromagnetic radiation in medical application for therapeutic uses such as treatment of skin disorders. For example, U.S. Patent No. 4,298,005 to Mutzhas describes a continuous ultraviolet lamp with cosmetic, photobiological, and photochemical applications. A treatment based on using the UV portion of the spectrum and its photochemical interaction with the skin is described. The power delivered to the skin using Mutzhas' lamp is described as 150W/m^2 , which does not have a significant effect on skin temperature.

In addition to prior art treatment involving UV light, lasers have been used for dermatological procedures, including Argon lasers, CO_2 lasers, Nd(Yag) lasers, Copper vapor lasers, ruby lasers and dye lasers. For example, U.S. Patent No. 4,829,262 to Furumoto, describes a method of constructing a dye laser used in dermatology applications. Two skin conditions which may be treated by laser radiation are external skin irregularities such as local differences in the pigmentation or structure of the skin, and vascular disorders lying deeper under the skin which cause a variety of skin abnormalities including port wine stains, telangiectasias, leg veins and cherry and spider angiomas. Laser treatment of these skin disorders generally includes localized heating of the treatment area by absorption of laser radiation. Heating the skin changes or corrects the skin

disorder and causes the full or partial disappearance of the skin abnormality.

Certain external disorders such as pigmented lesions can also be treated by heating the skin very fast to a high enough temperature to evaporate parts of the skin. Deeper-lying vascular disorders are more typically treated by heating the blood to a high enough temperature to cause it to coagulate. The disorder will then eventually disappear. To control the treatment depth a pulsed radiation source is often used. The depth the heat penetrates in the blood vessel is controlled by controlling the pulse width of the radiation source. The absorption and scattering coefficients of the skin also affect the heat penetration. These coefficients are a function of the constituents of skin and the wavelength of the radiation. Specifically, the absorption coefficient of light in the epidermis and dermis tends to be a slowly varying, monotonically decreasing function of wavelength. Thus, the wavelength of the light should be chosen so that the absorption coefficient is optimized for the particular skin condition and vessel size being treated.

The effectiveness of lasers for applications such as tattoo removal and removal of birth and age marks is diminished because lasers are monochromatic. A laser of a given wavelength may be effectively used to treat a first type of skin pigmentation disorder, but, if the specific wavelength of the laser is not absorbed efficiently by skin having a second type of disorder, it will be ineffective for the second type of skin disorder. Also, lasers are usually complicated, expensive to manufacture, large for the amount of power delivered, unreliable and difficult to maintain.

The wavelength of the light also affects vascular disorder treatment because blood content in the vicinity of the vascular disorders varies, and blood content affects the absorption coefficient of the treatment area. Oxyhemoglobin is the main chromophore which controls the optical properties of blood and has strong absorption bands in the visible region. More particularly, the strongest absorption peak of oxyhemoglobin occurs at 418nm and has a band-width of 60nm. Two additional absorption peaks with lower absorption coefficients occur at 542 and 577nm. The total band-width of these two peaks is on the order of 100nm. Additionally, light in the wavelength range of 500 to 600nm is desirable for the treatment of blood vessel disorders of the skin since it is absorbed by the blood and penetrates through the skin. Longer wavelengths up to 1000nm are also effective since they can penetrate deeper into the skin, heat the surrounding tissue and, if the pulse-width is long enough, contribute to heating the blood vessel by thermal conductivity. Also, longer wavelengths are effective for treatment of larger diameter vessels because the lower absorption coefficient is compensated for by the longer path of light in the vessel.

Accordingly, a wide band electromagnetic radiation source that covers the near UV and the visible portion of the spectrum would be desirable for treatment of ex-

ternal skin and vascular disorders. The overall range of wavelengths of the light source should be sufficient to optimize treatment for any of a number of applications. Such a therapeutic electromagnetic radiation device should also be capable of providing an optimal wavelength range within the overall range for the specific disorder being treated. The intensity of the light should be sufficient to cause the required thermal effect by raising the temperature of the treatment area to the required temperature. Also, the pulse-width should be variable over a wide enough range so as to achieve the optimal penetration depth for each application. Therefore, it is desirable to provide a light source having a wide range of wavelengths, which can be selected according to the required skin treatment, with a controlled pulse-width and a high enough energy density for application to the affected area.

Pulsed non-laser type light sources such as linear flashlamps provide these benefits. The intensity of the emitted light can be made high enough to achieve the required thermal effects. The pulse-width can be varied over a wide range so that control of thermal depth penetration can be accomplished. The typical spectrum covers the visible and ultraviolet range and the optical bands most effective for specific applications can be selected, or enhanced using fluorescent materials. Moreover, non-laser type light sources such as flashlamps are much simpler and easier to manufacture than lasers, are significantly less expensive for the same output power and have the potential of being more efficient and more reliable. They have a wide spectral range that can be optimized for a variety of specific skin treatment applications. These sources also have a pulse length that can be varied over a wide range which is critical for the different types of skin treatments.

In addition to being used for treating skin disorders, lasers have been used for invasive medical procedures such as lithotripsy and removal of blood vessel blockage. In such invasive procedures laser light is coupled to optical fibers and delivered through the fiber to the treatment area. In lithotripsy the fiber delivers light from a pulsed laser to a kidney or gallstone and the light interaction with the stone creates a shock wave which pulverizes the stone. To remove blood vessel blockage the light is coupled to the blockage by the fiber and disintegrates the blockage. In either case the shortcomings of lasers discussed above with respect to laser skin treatment are present. Accordingly, a treatment device for lithotripsy and blockage removal utilizing a flashlamp would be desirable.

To effectively treat an area the light from the source must be focussed on the treatment area. Coupling pulsed laser light into optical fibers in medicine is quite common. The prior art describes coupling isotropic incoherent point sources such as CW lamps into small optical fibers. For example, U. S. Patent No. 4,757,431, issued July 12, 1988, to Cross, et al. discloses a method for focusing incoherent point sources with small fila-

ments or an arc lamp with an electrode separation of 2mm into a small area. Point (or small) sources are relatively easy to focus without large losses in energy because of the small size of the source. Also, U. S. Patent No. 4,022,534, issued May 10, 1977, to Kishner discloses light produced by a flash tube and the collection of only a small portion of the light emitted by the tube into an optical fiber.

However, the large dimension of an extended source such as a flashlamp makes it difficult to focus large fractions of its energy into small areas. Coupling into optical fibers is even more difficult since not only must a high energy density be achieved, but the angular distribution of the light has to be such that trapping in the optical fiber can be accomplished. Thus, it is desirable to have a system for coupling the output of a high intensity, extended, pulsed light source into an optical fiber.

SUMMARY OF THE PRESENT INVENTION

According to a first embodiment of the invention a therapeutic treatment device comprises a housing and an incoherent light source, suitably a flashlamp, operable to provide a pulsed light output for treatment, disposed in the housing. The housing has an opening and is suitable for being disposed adjacent a skin treatment area. A reflector is mounted within the housing proximate the light source, and at least one optical filter is mounted proximate the opening in the housing. An iris is mounted coextensively with the opening. Power to the lamp is provided by a variable pulse width pulse forming circuit. Thus, the treatment device provides controlled density, filtered, pulsed light output through an opening in the housing to a skin area for treatment.

According to a second embodiment of the invention a method of treatment with light energy comprises the steps of providing a high power, pulsed light output from a non-laser, incoherent light source and directing the pulsed light output to a treatment area. The pulse width of the light output is controlled and focussed so that the power density of the light is controlled. Also, the light is filtered to control the spectrum of the light.

According to a third embodiment of the invention a coupler comprises an incoherent light source such as a toroidal flashlamp. A reflector is disposed around the incoherent light source and at least one optical fiber or light guide. The fiber has an end disposed within the reflector. This end collects the light from the circular lamp. In a similar coupling configuration fibers may be provided, along with a linear to circular fiber transfer unit disposed to receive light from the light source and provide light to the optical fibers. The reflector has an elliptical cross-section in a plane parallel to the axis of the linear flash tube, and the linear flash tube is located at one focus of the ellipse while the linear to circular transfer unit is located at the other focus of the ellipse.

BRIEF DESCRIPTION OF THE DRAWINGS

For a better understanding of the invention, reference is made to the accompanying drawings, in which like numerals designate corresponding elements or sections throughout, and in which:

Figure 1 is a cross-sectional view of an incoherent, pulsed light source skin treatment device;
 Figure 2 is a side view of the light source of Figure 1;
 Figure 3 is a schematic diagram of a pulse forming network with a variable pulse width for use with the skin treatment device of Figures 1 and 2;
 Figure 4 is a cross-sectional view of a coupler for coupling light from a toroidal flash tube into an optical fiber with a conical edge;
 Figure 5 is a side view of a toroidal flash tube;
 Figure 6 is a top view of a toroidal flash tube;
 Figure 7 shows the geometry for coupling into a conical section;
 Figure 8 is a cross-sectional view of a coupler for coupling light from a toroidal flash tube into an optical fiber with a flat edge;
 Figure 9 is a front sectional view of a coupler for coupling light from a linear flash tube into a circular fiber bundle;
 Figure 10 is a side sectional view of the coupler of Figure 9;
 Figure 11 is a front view of a coupler for coupling light from a linear flash tube into an optical fiber;
 Figure 12 is a front view of a coupler for coupling light from a linear flash tube into a doped optical fiber;
 Figure 13 is a schematic configuration of a gel skin interface with a transparent plate;
 Figure 14 shows an angular distribution of photons penetrating without using a gel;
 Figure 15 shows a light guide providing a large angular divergence;
 Figure 16 shows a light guide providing a narrow angular divergence;
 Figure 17 shows a spectra produced with a flashlamp current of 200 amps;
 Figure 18 shows a spectra produced with a flashlamp current of 200 amps; and
 Figure 19 shows a GTO driver circuit for a flashlamp.

In the various figures, like reference numerals are used to describe like components.

DETAILED DESCRIPTION OF THE PREFERRED EMBODIMENTS

Before explaining at least one embodiment of the invention in detail it is to be understood that the invention is not limited in its application to the details of construction and the arrangement of the components set forth in

the following description or illustrated in the drawings. The invention is capable of other embodiments or of being practiced or carried out in various ways. Also, it is to be understood that the phraseology and terminology employed herein is for the purpose of description and should not be regarded as limiting.

Referring now to Figures 1 and 2, cross-sectional and side views of an incoherent, pulsed light source skin treatment device 10 constructed and operated in accordance with the principles of the present invention are shown. The device 10 may be seen to include a housing 12, having an opening therein, a handle 13 (Figure 2 only), a light source 14 having an outer glass tube 15, an elliptical reflector 16, a set of optical filters 18, an iris 20 and a detector 22 (Figure 1 only).

Light source 14, which is mounted in housing 12, may be a typical incoherent light source such as a gas filled linear flashlamp Model No. L5568 available from ILC. The spectrum of light emitted by gas filled linear flashlamp 14 depends on current density, type of glass envelope material and gas mixture used in the tube. For large current densities (e.g., 3000 A/cm² or more) the spectrum is similar to a black body radiation spectrum. Typically, most of the energy is emitted in the 300 to 1000nm wavelength range.

To treat a skin (or visible) disorder a required light density on the skin must be delivered. This light density can be achieved with the focusing arrangement shown in Figures 1 and 2. Figure 1 shows a cross-section view of reflector 16, also mounted in housing 12. As shown in Figure 1, the cross-section of reflector 16 in a plane is perpendicular to the axis of flashlamp 14 is an ellipse. Linear flashlamp 14 is located at one focus of the ellipse and reflector 16 is positioned in such a way that the treatment area of skin 21 is located at the other focus. The arrangement shown is similar to focusing arrangements used with lasers and efficiently couples light from flashlamp 14 to the skin. This arrangement should not, however, be considered limiting. Elliptical reflector 16 may be a metallic reflector, typically polished aluminum which is an easily machinable reflector and has a very high reflectivity in the visible, and the UV range of the spectrum can be used. Other bare or coated metals can also be used for this purpose.

Optical and neutral density filters 18 are mounted in housing 12 near the treatment area and may be moved into the beam or out of the beam to control the spectrum and intensity of the light. Typically, 50 to 100nm bandwidth filters, as well as low cutoff filters in the visible and ultraviolet portions of the spectrum, are used. In some procedures it is desirable to use most of the spectrum, with only the UV portion being cut off. In other applications, mainly for deeper penetration, it is preferable to use narrower bandwidths. The bandwidth filters and the cutoff filters are readily available commercially.

Glass tube 15 is located coaxially with flashlamp 14 and has fluorescent material deposited on it. Glass tube

15 will typically be used for treatment of coagulation of blood vessels to optimize the energy efficiency of device 10. The fluorescent material can be chosen to absorb the UV portion of the spectrum of flashlamp 14 and generate light in the 500 to 650nm range that is optimized for absorption in the blood. Similar materials are coated on the inner walls of commercial fluorescent lamps. A typical material used to generate "warm" white light in fluorescent lamps has a conversion efficiency of 80%, has a peak emission wavelength of 570nm and has a bandwidth of 70nm and is useful for absorption in blood. The few millisecond decay time of these phosphors is consistent with long pulses that are required for the treatment of blood vessels.

Other shapes or configurations of flashlamp 14 such as circular, helical, short arc and multiple linear flashlamps may be used. Reflector 16 may have other designs such as parabolic or circular reflectors. The light source can also be used without a reflector and the required energy and power density may be achieved by locating light source 14 in close proximity to the treatment area.

Iris 20 is mounted in housing 12 between optical filters 18 and the treatment area and controls the length and the width of the exposed area, i.e. by collimating the output of flashlamp 14. The length of flashlamp 14 controls the maximum length that can be exposed. Typically an 8cm long (arc length) tube will be used and only the central 5cm of the tube is exposed. Using the central 5cm assures a high degree of uniformity of energy density in the exposed skin area. Thus, in this embodiment the iris 20 (also called a collimator) will enable exposure of skin areas of a maximum length of 5cm. The iris 20 may be closed to provide a minimum exposure length of one millimeter. Similarly, the width of the exposed skin area can be controlled in the range of 1 to 5mm for a 5mm wide flashlamp. Larger exposed areas can be easily achieved by using longer flash tubes or multiple tubes, and smaller exposure areas are obtainable with an iris that more completely collimates the beam. The present invention provides a larger exposure area compared to prior art lasers or point sources and is very effective in the coagulation of blood vessels since blood flow interruption over a longer section of the vessel is more effective in coagulating it. The larger area exposed simultaneously also reduces the required procedure time.

Detector 22 (Figure 1) is mounted outside housing 12 and monitors the light reflected from the skin. Detector 22 combined with optical filters 18 and neutral density filters can be used to achieve a quick estimate of the spectral reflection and absorption coefficients of the skin. This may be carried out at a low energy density level prior to the application of the main treatment pulse. Measurement of the optical properties of the skin prior to the application of the main pulse is useful to determine optimal treatment conditions. As stated above, the wide spectrum of the light emitted from the non-laser type

source enables investigation of the skin over a wide spectral range and choice of optimal treatment wavelengths.

In an alternative embodiment, detector 22 or a second detector system may be used for real-time temperature measurement of the skin during its exposure to the pulsed light source. This is useful for skin thermolysis applications with long pulses in which light is absorbed in the epidermis and dermis. When the external portion of the epidermis reaches too high a temperature, permanent scarring of the skin may result. Thus, the temperature of the skin should be measured. This can be realized using infra-red emission of the heated skin, to prevent over-exposure.

A typical real-time detector system would measure the infra-red emission of the skin at two specific wavelengths by using two detectors and filters. The ratio between the signals of the two detectors can be used to estimate the instantaneous skin temperature. The operation of the pulsed light source can be stopped if a preselected skin temperature is reached. This measurement is relatively easy since the temperature threshold for pulsed heating that may cause skin scarring is on the order of 50°C or more, which is easily measurable using infra-red emission.

The depth of heat penetration depends on the light absorption and scattering in the different layers of the skin and the thermal properties of the skin. Another important parameter is pulse-width. For a pulsed light source, the energy of which is absorbed in an infinitesimally thin layer, the depth of heat penetration (d) by thermal conductivity during the pulse can be written as shown in Equation 1:

$$(Eq. 1) \quad d = 4 [k\Delta t/C\rho]^{1/2}$$

where

k = heat conductivity of the material being illuminated;

Δt = the pulse-width of the light pulse;

C = the heat capacity of the material;

ρ = density of the material.

It is clear from Equation 1 that the depth of heat penetration can be controlled by the pulse-width of the light source. Thus, a variation of pulse-width in the range of 10^{-5} sec to 10^{-1} sec will result in a variation in the thermal penetration by a factor of 100.

Accordingly, the flashlamp 14 provides a pulse width of from 10^{-5} sec to 10^{-1} sec. For treatment of vascular disorders in which coagulation of blood vessels in the skin is the objective the pulse length is chosen to uniformly heat as much of the entire thickness of the vessel as possible to achieve efficient coagulation. Typical blood vessels that need to be treated in the skin have thicknesses in the range of 0.5mm. Thus, the op-

timal pulse-width, taking into account the thermal properties of blood, is on the order of 100msec. If shorter pulses are used, heat will still be conducted through the blood to cause coagulation, however, the instantaneous temperature of part of the blood in the vessel and surrounding tissue will be higher than the temperature required for coagulation and may cause unwanted damage.

For treatment of external skin disorders in which evaporation of the skin is the objective, a very short pulse-width is used to provide for very shallow thermal penetration of the skin. For example, a 10^{-5} sec pulse will penetrate (by thermal conductivity) a depth of the order of only 5 microns into the skin. Thus, only a thin layer of skin is heated, and a very high, instantaneous temperature is obtained so that the external mark on the skin is evaporated.

Figure 3 shows a variable pulse-width pulse forming circuit comprised of a plurality of individual pulse forming networks (PFN's) that create the variation in pulse-widths of flashlamp 14. The light pulse full width at half maximum (FWHM) of a flashlamp driven by a single element PFN with capacitance C and inductance L is approximately equal to:

$$(Eq. 2) \quad \Delta t \approx 2[LC]^{1/2}$$

Flashlamp 14 may be driven by three different PFN's, as shown in Figure 3. The relay contacts R1', R2' and R3' are used to select among three capacitors C1, C2 and C3 that are charged by the high voltage power supply. Relays R1, R2 and R3 are used to select the PFN that will be connected to flashlamp 14. The high voltage switches S1, S2 and S3 are used to discharge the energy stored in the capacitor of the PFN into flashlamp 14. In one embodiment L1, L2 and L3 have values of 100mH, 1mH and 5mH, respectively, and C1, C2 and C3 have values of 100mF, 1mF and 10mF, respectively.

In addition to the possibility of firing each PFN separately, which generates the basic variability in pulse-width, additional variation can be achieved by firing PFN's sequentially. If, for example, two PFN's having pulse-width Δt_1 and Δt_2 are fired, so that the second PFN is fired after the first pulse has decayed to half of its amplitude, then an effective light pulse-width of this operation of the system will be given by the relation: $\Delta t \approx \Delta t_1 + \Delta t_2$. The charging power supply typically has a voltage range of 500V to 5kV. The relays should therefore be high voltage relays that can isolate these voltages reliably. The switches S are capable of carrying the current of flashlamp 14 and to isolate the reverse high voltage generated if the PFNs are sequentially fired. Solid-state switches, vacuum switches or gas switches can be used for this purpose.

A simmer power supply (not shown in Figure 3) may be used to keep the flashlamp in a low current conducting mode. Other configurations can be used to achieve

pulse-width variation, such as the use of a single PFN and a crowbar switch, or use of a switch with closing and opening capabilities.

Typically, for operation of flashlamp 14 with an electrical pulse-width of 1 to 10msec, a linear electrical energy density input of 100 to 300J/cm can be used. An energy density of 30 to 100J/cm² can be achieved on the skin for a typical flashlamp bore diameter of 5mm. The use of a 500 to 650nm bandwidth transmits 20% of the incident energy. Thus, energy densities on the skin of 6 to 20J/cm² are achieved. The incorporation of the fluorescent material will further extend the output radiation in the desired range, enabling the same exposure of the skin with a lower energy input into flashlamp 14.

Pulsed laser skin treatment shows that energy densities in the range of 0.5 to 10J/cm² with pulse-widths in the range of 0.5msec are generally effective for treating vascular related skin disorders. This range of parameters falls in the range of operation of pulsed non-laser type light sources such as the linear flashlamp. A few steps of neutral density glass filters 18 can also be used to control the energy density on the skin.

For external disorders a typical pulse-width of 5 microsecond is used. A 20J/cm electrical energy density input into a 5mm bore flashlamp results in an energy density on the skin of 10J/cm². Cutting off the hard UV portion of the spectrum results in 90% energy transmission, or skin exposure to an energy density of close to 10 J/cm². This energy density is high enough to evaporate external marks on the skin.

Device 10 can be provided as two units: a lightweight unit held by a physician using handle 13, with the hand-held unit containing flashlamp 14, filters 18 and iris 20 that together control the spectrum and the size of the exposed area and the detectors that measure the reflectivity and the instantaneous skin temperature. The power supply, the PFN's and the electrical controls are contained in a separate box (not shown) that is connected to the hand-held unit via a flexible cable. This enables ease of operation and easy access to the areas of the skin that need to be treated.

The invention has thus far been described in conjunction with skin treatment. However, using a flashlamp rather than a laser in invasive treatments provides advantages as well. Procedures such as lithotripsy or removal of blood vessel blockage may be performed with a flashlamp. Such a device may be similar to that shown in Figures 1 and 2, and may use the electronics of Figure 3 to produce the flash. However, to properly couple the light to an optical fiber a number of couplers 40, 80 and 90 are shown in Figures 4 and 8-10, respectively.

Coupler 40 includes an optical source of high intensity incoherent and isotropic pulsed light such as a linear flash tube 42, a light reflector 44 which delivers the light energy to an optical fiber 46. The latter has a generally conical edge in the embodiment of Figure 4. Optical fiber 46 transfers the light from light collection system 44 to the treatment area. In general, coupler 40 couples

pulsed light from a flash tube into an optical fiber and has applications in medical, industrial and domestic areas.

For example, coupler 40 may be used in material processing to rapidly heat or ablate a portion of a material being processed, or to induce a photo-chemical process. Alternatively, coupler 40 may be used in a photography application to provide a flash for picture taking. Using such a coupler would allow the flash bulb to be located inside the camera, with the light transmitted to outside the camera using an optical fiber. As one skilled in the art should recognize coupler 40 allows the use of incoherent light in many applications that coherent or incoherent light has been used in the past.

To provide for coupling the light to an optical fiber, flash tube 42 has a toroidal shape, shown in Figures 5 and 6, and is disposed inside reflector 44. In addition to the toroidal shape other shapes, such as a continuous helix, may be used for flash tube 42. However, a helical tube is more difficult to manufacture than a toroidal tube. Referring now to Figure 6, flash tube 42 is generally in the shape of a torus, but is not a perfect torus since the electrodes located at the end of the torus have to be connected to the power source. This does not create a significant disturbance in the circular shape of flash tube 42, since the connection to the electrodes can be made quite small.

Reflector 44 collects and concentrates the light, and has a cross-section of substantially an ellipse, in a plane perpendicular to the minor axis of the toroidal flash tube 42. The major axis of this ellipse preferably forms a small angle with the major axis of toroidal lamp 42. The exact value of the angle between the ellipse axis and the main axis of lamp 42 depends on the Numerical Aperture (NA) of the optical fiber. The toroidal flash tube is positioned so that its minor axis coincides with the focus of the ellipse. The other focus of the ellipse is at the edge of optical fiber 46. Reflector 44 may be machined from metal with the inner surfaces polished for good reflectivity. Aluminum is a very good reflector with high reflectivity in the visible and ultraviolet wavelengths, and it may be used for this purpose. The reflector can be machined in one piece and then cut along a surface perpendicular to the main axis of the device. This will enable integration of the toroidal flash tube into the device.

As shown in Figure 4, the edge of optical fiber 46 is a cone with a small opening angle, so that the total area of the fiber exposed to the light from the flash tube is increased. Referring now to Figure 7 the geometry for coupling light into a conical tip is shown. It is assumed here that the light comes from a region in space with a refractive index of n_2 and that the conical section of the fiber (as well as the rest of the fiber core) has a refractive index of n_1 .

Not all the light rays hitting the cone are trapped in it. For light rays that propagate in a plane that contains the major axis of the system, a condition can be derived for the angle of a ray that will be trapped and absorbed

in the fiber. This condition is shown in Equation 3.

$$\sin(\mu_{\text{criti}}) = \cos(\beta) - [n_1^2/n_2^2 - 1]^{1/2} \sin(\beta) \quad (\text{Eq. 3})$$

Light will be trapped in the conical portion of the optical fiber if the incidence angle μ is larger than μ_{criti} calculated from Equation 3. Trapping is possible only if $n_1 > n_2$. If the medium outside of the fiber is air, $n_2 = 1$. Not all of the light trapped in the conical section of the fiber will also be trapped in the straight portion of the fiber if a fiber with a core and a cladding is used. If a fiber with a core and no cladding is used (air cladding), then all the rays captured in the conical section of the fiber will also be trapped in the straight section of the fiber.

The configuration shown in Figure 4 can also be used with a fluid filling the volume between the reflector and the optical fiber. A very convenient fluid for this purpose may be water. Water is also very effective in cooling the flashlamp if high repetition rate pulses are used. The presence of a fluid reduces the losses that are associated with glass to air transitions, such as the transition between the flashlamp envelope material and air. If a fluid is used in the reflector volume, then its refractive index can be chosen such that all the rays trapped in the conical section are also trapped in the fiber, even if core/cladding fibers are used.

Another way of configuring the fiber in the reflector is by using a fiber with a flat edge. This configuration is shown in Figure 8 and has trapping efficiency very close to the trapping efficiency of the conical edge. Many other shapes of the fiber edge, such as spherical shapes, can also be used. The configuration of the fiber edge also has an effect on the distribution of the light on the exit side of the fiber and it can be chosen in accordance with the specific application of the device.

The device may be used with a variety of optical fibers. Single, or a small number of millimeter or sub-millimeter diameter fibers, will typically be used in invasive medical applications. In other applications, particularly in industrial and domestic applications, it may be preferable to use a fiber having a larger diameter, or a larger bundle of fibers, or a light guide.

According to one embodiment flexible or rigid light guides are used to couple the light to the treatment area. Flexible light guides made from a bundle of quartz or other glass fibers that are fused together by heat at the edge of the bundles. The bundles may be circular,

rectangular, or any other useful shape. Rigid light guides may be made from quartz, acrylic, glass, or other materials having a high degree of transparency. The material is generally highly polished on all sides.

A typical cross section of a circular light guide useful for therapeutic treatment is one mm to ten mm in diameter. Alternatively, a rectangular light guide may be used having typical dimensions of 3 mm by 10 mm to 30 mm by 100 mm. In either case the length may be 20 to 300

mmm, or as needed for the specific application.

According to another alternative embodiment a rectangular light guide is used to more efficiently couple the light. The rectangular light guide is chosen to have a shape that matches a rectangular linear flashlamp and to match the shape of the vessel being treated.

The light guides described above may be used in another alternative embodiment to control the spectrum of light delivered to the treatment area. Spectral control can be achieved by making the light guide from a material that had an absorbing dye dissolved therein. Thus, light transmitted by the light guide will have a spectrum in as determined by the absorbing dye. Alternatively, a flat, discrete filter may be added to one end (preferably the input end) of the light guide. Both of these filters are absorbing filters. The inventors have found that absorbing filters produced by Schott, having Model Nos. OG515, OG550, OG570, and OG590 have suitable characteristics.

Additionally, interference filters or reflective coatings on the light guide may be used by applying a proper optical coating to the light guide. Again, a single discrete interference filter could also be used. Additionally, combinations of the various filters described herein, or other filters, may be used. The use of the filters described here may render the use of the filters described earlier with reference to Figure 1 redundant.

An alternative embodiment entails the use of application specific light guides. In this way the spectra of light for various treatments can be easily controlled. According to this alternative each type of treatment will be performed with a specific light guide.

The optical properties of the light guide will be chosen to optimize the particular treatment. The wavelengths below are particularly useful for the respective treatments:

arteries less than 0.1 mm in diameter - 520-650nm
veins less than 0.1 mm in diameter - 520-700nm
vessels between 0.1 and 1.0 mm in diameter - 550-1000nm
larger vessels - 600-1000nm

In each case if the skin is darker (higher pigmentation) longer wavelengths on the lower cut-off portion of the spectrum should be used.

Multiple spectra may be used for optimal penetration. This may be accomplished by illuminating with a few pulses, each having a different spectrum. For example, the first pulse can have a spectrum that is highly absorbed in blood. This pulse will coagulate the blood, thereby changing the optical properties of the blood, making it more absorbing in another wavelength range (preferably longer). A second pulse will be more efficiently absorbed since the blood absorbs energy of a greater wavelength range. This principle may be used with lasers or other light sources as well.

In addition to the features of the light guides dis-

cussed above, a light guide is used, in one alternative embodiment, to control the angular distribution of the light rays impinging on the skin. Light that impinges on the skin at large angles (relative to the perpendicular) will not penetrate very deeply into the tissue. Conversely, light that impinges perpendicularly to the skin will have a deeper penetration. Thus, it is desirable to provide a distribution of light rays that has a relatively wide angular divergence when the treatment requires shallow penetration. Alternatively, a narrow divergence is preferable for treatment requiring deep penetration is desired. Some treatment might require both shallow and deep penetration.

Figure 15 shows a light guide 115 having an exit beam with a greater angular divergence than that of the entrance beam. As shown in Figure 15, a beam 116 enters light guide 115 at a small angle, relative to the axis of light guide 115. When beam 501 exits light guide 115, the angle, relative to the axis, is much greater. The tapered shape of light guide 115 enhances this divergence.

Figure 16 shows a straight light guide 118 that maintains the angular distribution of the rays of light that enter into it. A beam 119 is shown entering and exiting light guide 118 with the same angle, relative to the axis of coupler 601. Alternate use of both light guides 115 and 118 can achieve the narrow and deep penetration discussed above. Alternatively, the user can select the type of coupler according to the depth of penetration needed for the treatment being performed.

Figures 9 and 10 show a coupler 90 for coupling a linear flash tube 92 through a linear to circular fiber transfer unit 94 to a fiber bundle 96. A reflector 98 has an elliptical cross-section, shown in Figure 10, in a plane parallel to the axis of linear flash tube 92 in this embodiment. Tube 92 is located on one focus of the ellipse while the linear side of linear to circular bundle converter 94 is located at the other focus of the ellipse. This configuration is relatively simple to manufacture and commercially available linear to circular converters such as 25-004-4 available from General Fiber Optics may be used. This configuration is particularly useful for larger exposure areas of the fiber, or for flash illumination purposes.

The energy and power densities that can be achieved by this invention are high enough to get the desired effects in surface treatment or medical applications. For the embodiment shown in Figure 4 the total energy and power densities can be estimated as follows. For a typical toroidal lamp with a 4mm bore diameter and a major diameter of 3.3cm an electrical linear energy density input of 10J/cm into the lamp can be used with a 5μsec pulse width. The light output of the lamp will be 5 to 6J/cm for optimal electrical operating conditions. For the reflector shown in Figure 4, 50% of the light generated in the lamp will reach the lower focus. Thus, a total energy flux on the focus of 25 to 30J may be obtained. For embodiments shown in Figure 4 or Fig-

ure 8 the total cross-section area of reflector at the focal plane has a cross-section of 0.8cm^2 . Energy densities on the order of 30 to 40J/cm^2 at the entrance to the fiber should be attained with this cross-section. This corresponds to power densities of 5 to 10MW/cm^2 , which are the typical power densities used in medical or material processing applications.

For longer pulses, higher linear electrical energy densities into the lamp can be used. For a 1msec pulse to the flash tube a linear electrical energy density of 100J/cm can be used. The corresponding energy density at the focal area would be up to 300J/cm^2 . Such energy densities are very effective in industrial cleaning and processing applications as well as in medical applications.

Alternative embodiments for coupling the optical fiber to an extended light source such as a linear flashlamp are shown in Figures 11 and 12. In the embodiment of Figure 11 an optical fiber 101 is wound around a lamp 102 and a lamp envelope 103. Some of the light that is produced by the light source is coupled into the fiber. If the light rays are propagating in the direction that is trapped by the fiber then this light will propagate in the fiber and it can be used at a fiber output 104. One limitation of this configuration is the fact that most of the light emitted by lamp 103 travels in a direction perpendicular to the surface of lamp 103 and cannot be trapped in fiber 101.

The embodiment shown in Figure 12 overcomes this problem. A doped optical fiber 105 is wound around lamp 102 and envelope 103, rather than an undoped fiber such as fiber 101 of Figure 11. The dopant is a fluorescent material which is excited by the radiation emanating from lamp 102 and radiates light inside the fiber. This light is radiated omnidirectionally and the part of it that is within the critical angle of fiber 105 is trapped and propagates through the fiber and can be used at fiber output 104. The angle of light that is trapped in the fiber is the critical angle of the material from which the optical fiber or optical wave guide is made. For a fiber (or optical wave guide) in air this angle is given by $\sin \alpha = 1/n$.

Typically for glass or other transparent materials $n = 1.5$ and $\alpha = 41.8^\circ$. This corresponds to a trapping efficiency of more than 10% of the light emitted by fluorescence inside the fiber. If we assume a 50% efficiency of the fluorescence process we find out that more than 5% of the light produced by the lamp is trapped and propagated down the fiber. For example, a $4"$ lamp with a linear electrical energy input of 300J/inch and 50% electrical to light conversion efficiency would couple 2.5% of its electrical energy into the fiber. This corresponds, for the $4"$ lamp case to a total light energy of 30J of light. This embodiment has the additional advantage of transferring the wavelength emitted by the lamp to a wavelength that may be more useful in some of the therapeutic or processing applications mentioned before. Thus, fluorescent material doped in the fiber can be chosen in accordance with an emission wavelength

determined by the specific application of the device.

One alternative embodiment includes the use of a gel to couple the light to the skin. This alternative reduces heating of the outer layer of the skin (the epidermis and upper layers of the dermis). The gel is preferably a high viscosity water based gel and is applied to the skin before treatment, although other gels that are not necessarily water based may be used. A gel having a relatively high heat capacity and thermal conductivity, such as a water based gel, is preferable to enable cooling of the outer skin (the epidermis in particular). Transparency is also desirable because during treatment light passes through the transparent gel and reaches the skin.

Referring now to Figure 13, a gel 110 is applied to the skin 21 prior to the treatment. A flat layer of gel on top of the skin is used since irregularities in the upper layer of the gel through which the light passes may cause scattering of the light and reduce its penetration into the skin. In order to achieve this flatness a solid, transparent, flat piece 111 may be applied on top of the skin. The configuration is shown schematically in Figure 13. The transparent plate can be made out of glass or other transparent materials. Either the flashlamp housing or the light guides discussed above may be placed in direct contact with the transparent plate.

The configuration of Figure 13 has the advantage of reducing the scattering of light (represented by arrows 113) that enters into the skin due to irregularities in the surface of the skin. The skin has an index of refraction that is larger than that of the air. As a result, any photon that impinges on the air skin interface is deflected if it does not hit the skin at an incidence angle of 0° . Since the surface of the skin is irregular the angular distribution of the skin increases. This is shown schematically in Figure 14.

The use of gel addresses this problem since the gel can fill irregular voids that are created by the skin structure. The transparent plate that covers the gel and the gel itself will preferably have an index of refraction that is close to that of the skin. This is relatively easy since the index of refraction of the skin is of the order of 1.4 in the visible and the near infrared. Most glasses and transparent plastics have indices of refraction that are of the order of 1.5 which is close enough. The index of refraction of water is of the order of 1.34 in this range. Water based gels will have similar indices of refraction. The index can be increased by proper additives. The plate and gel thus act as a flat surface for the light to impinge upon. Because the gel and plate have an index of refraction close to that of the skin there is very little scattering at the gel-plate and gel-skin interfaces.

The use of a gel has been experimentally successful in the treatment of leg veins and other benign vascular lesions of the skin. The treatments were carried out with the flashlamp described above. However, in alternative embodiments a different incoherent source, or a coherent source, may be used.

During operation light is typically applied to the skin

in a sequence of three pulses with short delays between the pulses. This mode of operation is used in order to take advantage of the faster cooling of the superficial, thin (less than 0.1mm thick) epidermis compared to the larger and deeper vessels typical of leg veins. The gel in contact with the skin cools the epidermis during the waiting period between the pulses. This cooling reduces significantly the damage to the epidermis.

In accordance with the invention, light is applied to the treated area in either a long pulse or in a sequence of pulses separated by a delay. The delay and/or pulse length is preferably controlled by the operator to provide enough heat to accomplish the desired treatment but not enough heat to damage the skin.

This concept was tested with large and deep vessels (of the order of 2mm in diameter and 2mm deep). A thin layer of commercial water based ultrasound gel (1 to 2mm thick, "Aqua clear" gel made by Parker USA) was applied on the skin. A 1mm thin glass window was used to generate a flat layer of the gel. The light from the device passed through the thin glass and the gel and into the skin. Care was taken to assure that no air bubbles exist in the gel. This configuration was tested with photon fluences of 30 to 50J/cm². Coagulation and clearance of the vessels was obtained without causing damage to the skin. This is contrary to similar trials in which gel was not used and in which fluences of 20J/cm² with the same pulse structure caused burns of the skin.

The epidermis has a thickness of approximately 0.1 mm and a cooling time of about 5 msec. Thus, to avoid burning delays greater than 5 msec are used.

In another alternative embodiment the spectrum of the light used for treatment is controlled by controlling the voltage and/or current applied to the flashlamp. As is well known in the art, the spectrum of light produced by a flashlamp is dependent on the voltage and current provided to the flashlamp. According to this embodiment the input voltage and current is selected to provide a desired treatment spectrum. The appropriate voltage and currents may be determined experimentally for each flashlamp used. For example, a flashlamp current of 200 amps produced the spectra shown in Figure 17. Similarly, the spectra of Figure 18 was produced using a flashlamp current of 380 amps. The spectra of Figure 17 shows a significant enhancement in the wavelength range of 800-1000nm. Such a spectra is particularly useful for treatment of large vessels.

The different currents and voltages used to control the output spectra may be obtained using a group or bank of capacitors that are capable of being connected in either series or parallel as part of the power source for the flashlamp. A series connection will provide a relatively high voltage and high current, thereby producing a spectra having energy in a shorter wavelength, such as 500-650 nm. Such a series connection will be more appropriate for generating shorter pulses (1 to 10 msec, e.g.) useful for treatment of smaller vessels.

A parallel connection provides a lower current and voltage, and thus produces an output spectra of a longer wavelength, such as 700-1000 nm. Such a spectra is more appropriate for treatment of larger vessels and is suitable for producing longer pulses (in the range of 10-50 msec, e.g.). The selection of series or parallel connections may be done using a relay or sets of relays.

In one alternative embodiment the pulse forming network of Figure 3 is replaced by a GTO driver circuit 121, such as that shown in Figure 19. The driver circuit of Figure 19 uses a switch capable of being turned both on and off to control the application of power to the flashlamp. While this alternative embodiment will be described with respect to a GTO being used as the switch, other switches capable of being turned both on and off, such as IGBTs, may also be used.

Referring now to Figure 19, driver circuit 121 includes a high voltage source 122, a capacitor bank C5, an inductor L5, a diode D5, a switch GTO1, a diode D6, a diode D7, a resistor R5, a capacitor C6, a GTO trigger generator TR1, a resistor R7, a capacitor C7 and a flashtube trigger generator TR2. These components are connected to flashlamp 14 and serve to provide the power pulses to flashlamp 14. The duration and timing of the pulses are provided in accordance with the description herein. Driver 121 operates in the manner described below.

High voltage source 122 is connected across capacitor bank C5, and charges capacitor bank C5 to a voltage suitable for application to flashlamp 14. Capacitor bank C5 may be comprised of one or more capacitors, and may be configured in the manner described above.

Prior to illumination of flashlamp 14 flashtube trigger generator TR2 breaks down flashlamp 14 and creates a relatively low impedance channel therein. After the flashlamp breaks down, capacitor C7 dumps current into flashlamp 14, further creating a low impedance channel in flashlamp 14. In this manner a pre-discharge is provided that prepares flashlamp 14 for the power pulse. Capacitor C7 provides a small amount of current, relative to capacitor bank C5. Alternatively, driver circuit 121 may operate in a simmer mode, wherein the pre-discharge is not necessary.

Thereafter, switch GTO1 is turned on via a pulse from GTO trigger generator TR1, completing the circuit between flashlamp 14 and capacitor bank C5. Thus, capacitor bank C5 discharges through flashlamp 14. An inductor L5 may be provided to control the rise time of the current through flashlamp 14. Inductor L5 may include an inherent resistive component, not shown.

After a length of time determined by the desired pulse width has passed, GTO trigger generator TR1 provides a pulse to switch GTO1, turning it off. A control circuit determines the timing of the trigger pulses and provides them in accordance with the desired pulse widths and delays.

A snubber circuit comprised of diode D6, resistor

R5, and a capacitor C6 is provided for switch GTO1. Also, diodes D5 and D7 are provided to protect switch GTO1 from reverse voltages. Resistor R7 is provided in parallel with flashlamp 14 to measure the leakage current of switch GTO1, which can in turn be used to make sure that switch GTO1 is operating properly.

A possible addition to driver circuit 121 is to provide an SCR or other switch in parallel with capacitor bank C5. This allows the discharge or resetting of capacitor bank C5 without turning on switch GTO1. Other modifications may be made, such as providing the circuit with a serial trigger, rather than the parallel trigger shown. Another modification is to use the driver circuit with a laser rather than flashlamp 14.

Proper use of pulse widths and delays can aid in avoiding burning the epidermis. The epidermis has a cooling time of about 5 msec, while large vessels have a longer cooling time (a 1 mm vessel has a cooling time of about 300 msec). Thus, during a pulse of duration longer than 5 msec the epidermis can cool down but the vessel will not. For example, for treatment of a large vessel (such as one having a diameter of about 1mm) a pulse of 100 msec will allow the skin to cool, but the vessel will not cool.

The same effect may be achieved using trains of pulses. This is useful when it is not practical to provide a single long pulse to the flashlamp. The delays between pulses are selected to allow the skin to cool, but to be too short for the vessel to cool. Thus, larger vessels can be treated with longer delays because they have greater cooling times. Small vessels cool quickly and long delays are not effective. However, they also need less energy and can be treated effectively in a single pulse.

Typical delay times are in the range of 20 msec to 500 msec. More specifically, delays of between 100-500 msec are effective for vessels larger than 1 mm in diameter. Delays of between 20-100 msec are effective for vessels between 0.5 and 1 mm in diameter. Delays of between 10-50 msec are effective for vessels between 0.1 and 0.5 mm in diameter. A single pulse having a width in the range of 1 msec to 20 msec is effective for vessels less than 0.1 mm diameter.

Additionally, delays should be selected according to skin pigmentation. Darker skin absorbs more energy and needs more time to cool; thus longer delays are needed. Lighter skin absorbs less energy and can accommodate shorter delays.

It has been found that multiple pulses avoids "purpura" or the explosion of small vessels in or close to the skin. The use of pulses to avoid burning and provide cooling will be effective for light provided by lasers or other sources as well.

Another alternative embodiment includes the use of a microprocessor or personal computer to control the flashlamp. The microprocessor can be used to provide the timing functions and prompt the trigger signals described above. Additionally, in one embodiment the microprocessor includes a user interface, such as a display

screen and keyboard, buttons, mouse, or other input device. The microprocessors have information stored therein that aids in the selection of treatment parameters.

For example, if the condition being treated is a port wine stains skin type III, the physician inputs that condition into the microprocessor. The microprocessor responds with suggested treatment parameters, such as using a 570nm cut-off filter, a double pulse with a delay of 50 msec and a fluence of 55 J/cm². The physician can alter these suggested parameters, but need not refer back to operating guidelines for suggested parameters. This alternative may be used with light sources other than a flashlamp, such as UV or a pulsed laser.

These output parameters are shown on a display such as a screen or printer, and include the outputs discussed below. This will aid the physician in determining the proper treatment and in learning how to effectively use such devices. In one embodiment the microprocessor output on the display shows a simulation of interaction of light with skin and vascular lesions, oxygen concentration and temperature distribution in malignant tissue being illuminated for the purpose of cancer by a flashlamp, or processes occurring in skin resurfacing using infrared lasers or other sources.

A program within the microprocessor (or alternatively an analog circuit) models interaction of light with tissue and vessels. Many programs may be used to carry out the modeling, and in the preferred embodiment the following input parameters are used: light source type (flashlamp or pulsed laser e.g.); number of output curves (1-4 e.g.); skin type; vessel diameter and depth; blood type (oxy or deoxy - hemoglobin); pulse duration; delay between pulses; energy fluence; type of filter; short or long pulse mode; is a gel being used and its temperature. For a pulsed laser the wavelength is an input (400-1064 nm e.g.).

The microprocessor and the screen show the following information in one embodiment: temperature distribution in the tissue and in the vessel at the end of treatment; graphs of up to four curves to compare different light sources or treatment regime. Alternatively, the outputs could be printed rather than shown on a screen.

One skilled in the art will recognize that many microprocessor routines may be used to implement the invention. The routines may model the interactions in many ways, and one such model for single dimensional light interaction with a tissue uses the following empirical expression for fluence:

$$F=F(0)\exp(-x/d)$$

where $d = 1/\mu_{eff}$
and

$$\mu_{eff} = [3\mu_a(\mu_a + \mu_s(1-g))]^{1/2}$$

Where μ_a is an absorption coefficient of dermis, μ_s is a scattering coefficient of dermis, and g is the anisotropy factor which is defined as average cosine of scattering angle for one scattering event.

F(0) was calculated in accordance with *Diffusion of Light in Turbid Material*, A. Ishimaru, Applied Optics, 1989 Vol. 28 No. 12, pp 2210-2215, but an empirical correction depending on wavelength is added:

$$F(0) = F_0(640/W)^{1/4}$$

where W is a wavelength.

Light relaxation time in the tissue is significantly less than the temperature relaxation time and light pulse durations used for treatment of skin lesions, therefore a stationary model for description of light penetration into the tissue was used. Ishimaru's hydrodynamic model is suitable for calculating of $F(0)$. Accordingly to this model, the diffuse energy fluence rate ψ_d satisfies the following diffusion equation:

$$(\nabla^2 - \kappa^2) \psi_d = -Q$$

$$Q = 3\gamma_s (\gamma_T + g\gamma_a) F_0 \exp(-\tau)$$

$$\kappa^2 = 3\gamma_a \gamma_{tr}$$

$$\gamma_{tr} = \gamma_s(1 - g) + \gamma_a$$

$$\gamma_t = \gamma_a + \gamma_s$$

$$\tau = \int \gamma_t ds$$

where F_0 is the incident energy beam. Scattering and absorption coefficients are functions of wavelength. The total energy fluence rate is given by

$$\psi_t = \psi_d + \psi_c$$

$$\psi_c = F_0 \exp(-\tau).$$

This equation was calculated numerically with the corresponding boundary conditions. The boundary condition for ψ_d at the surface illuminated by the incident intensity is

$$\psi_d + \frac{2}{3\gamma_{tr}} \psi_d + \frac{2\gamma_s g F_0}{\gamma_{tr}} = 0$$

Temperature distribution behavior in the tissue is described by solution of the 1-D heat conductivity equation in planar geometry for near epidermis area

$$\rho c \frac{\partial T}{\partial t} = \lambda \frac{\partial^2 T}{\partial x^2} + \gamma_a \psi_t$$

with initial and boundary conditions

$$T_{T=0} = 36$$

$$\frac{\partial T}{\partial x} \Big|_{x=0} = 0$$

Here ρ is the density of tissue, c is the specific heat and λ is the heat conductivity coefficient. The thermal properties of water were assumed for the thermal properties of tissue.

A heat conductivity equation is calculated in cylindrical geometry for near vessel area, where the center of the cylinder was chosen as point with maximal temperature.

$$\rho c \frac{\partial T}{\partial t} = \lambda \frac{\partial^2 T}{\partial r^2} + \frac{1}{r} \frac{\partial T}{\partial r} + \gamma_a \psi_t$$

As one skilled in the art should recognize other models may be used as well.

The microprocessor or personal computer can also be used to create and store patient information in a database. Thus, past treatment information such as condition being treated, treatment parameters, number of treatments, etc. is stored and may be recalled when the patient is again treated. This aids in providing the proper treatment to the patient. Additionally, the database may include photographs of the patient's condition before and after each treatment. Again, this aids in record keeping and determining what treatments are most successful for given conditions.

In addition to the treatments described above the devices and methods described herein may be used to treat other conditions. For example, psoriasis and warts have been successfully treated. Similarly, skin rejuvenation (treating wrinkles) should be effective. The inventor further contemplates using this invention to treat hemorrhoids, throat lesions, and gynecological problems associated with vascular malformations.

Thus, it should be apparent that there has been provided in accordance with the present invention a flash-lamp and coupler that fully satisfy the objectives and advantages set forth above. Although the invention has been described in conjunction with specific embodiments thereof, it is evident that many alternatives, modifications and variations will be apparent to those skilled

in the art. Accordingly, it is intended to embrace all such alternatives, modifications and variations that fall within the spirit and broad scope of the appended claims.

Claims

1. A therapeutic treatment device comprising:

a light source operable to provide an incoherent pulsed light output for treatment, wherein the source includes control inputs;
 a microprocessor having control outputs connected to the control inputs of the source, display outputs, and an input interface;
 a display connected to the microprocessor display outputs;
 a user interface connected to the input interface;
 wherein the microprocessor includes means for receiving input parameters related to the treatment and means for providing output parameters related to the treatment; and
 means for causing the display to display the output parameters.

2. The therapeutic treatment device of claim 1, wherein the microprocessor is coupled to the user interface to receive information about the patient being treated.
3. The therapeutic treatment device of claim 2, wherein the information about the patient includes the condition of the patient being treated and the patient's treatment parameters.
4. The therapeutic treatment device of claim 3, wherein the patient information received by the microprocessor includes a picture of the patient's condition.
5. The therapeutic treatment device of claim 4, wherein the picture is of the patient's condition prior to treatment.
6. The therapeutic treatment device of claim 4, wherein the picture is of the patient's condition after treatment.
7. The therapeutic treatment device of claim 1, wherein the microprocessor is coupled to the display to generate visual indicia representative of the temperature distribution in a patient's tissue.
8. The therapeutic treatment device of claim 7, wherein the microprocessor is coupled to the display to generate visual indicia representative of the temperature in the patient's blood vessel at the end of treatment.

9. The therapeutic treatment device of claim 1, wherein the microprocessor is coupled to the display to generate visual indicia representative of suggested treatment parameters.

10. The therapeutic treatment device of claim 9, wherein the microprocessor is coupled to the display to generate visual indicia representative of a suggested filter to be placed between the incoherent light source and a patient.

11. The therapeutic treatment device of claim 9, wherein the microprocessor is coupled to the display to generate visual indicia representative of a suggested number of incoherent light pulses.

12. The therapeutic treatment device of claim 1, wherein the microprocessor is coupled to the display to generate visual indicia representative of a suggested time delay between each pulse of the number of pulses.

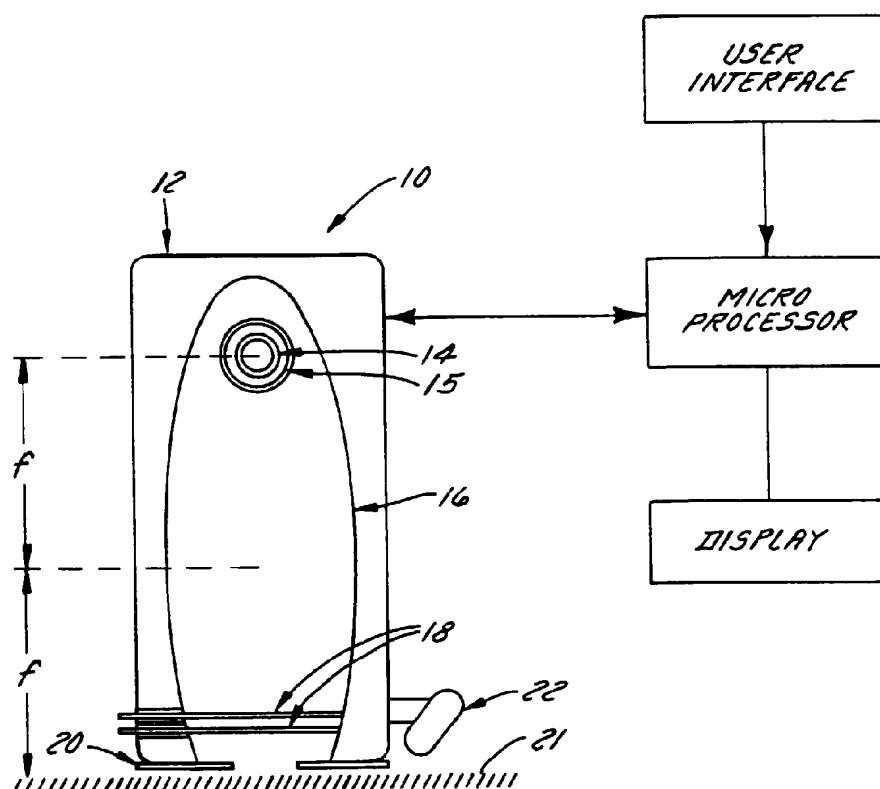


FIG. 1

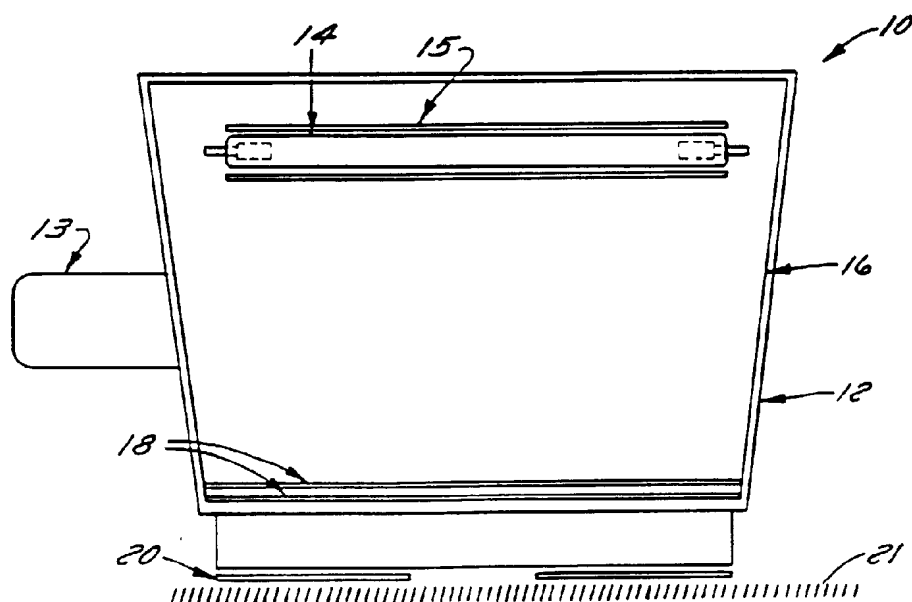


FIG. 2

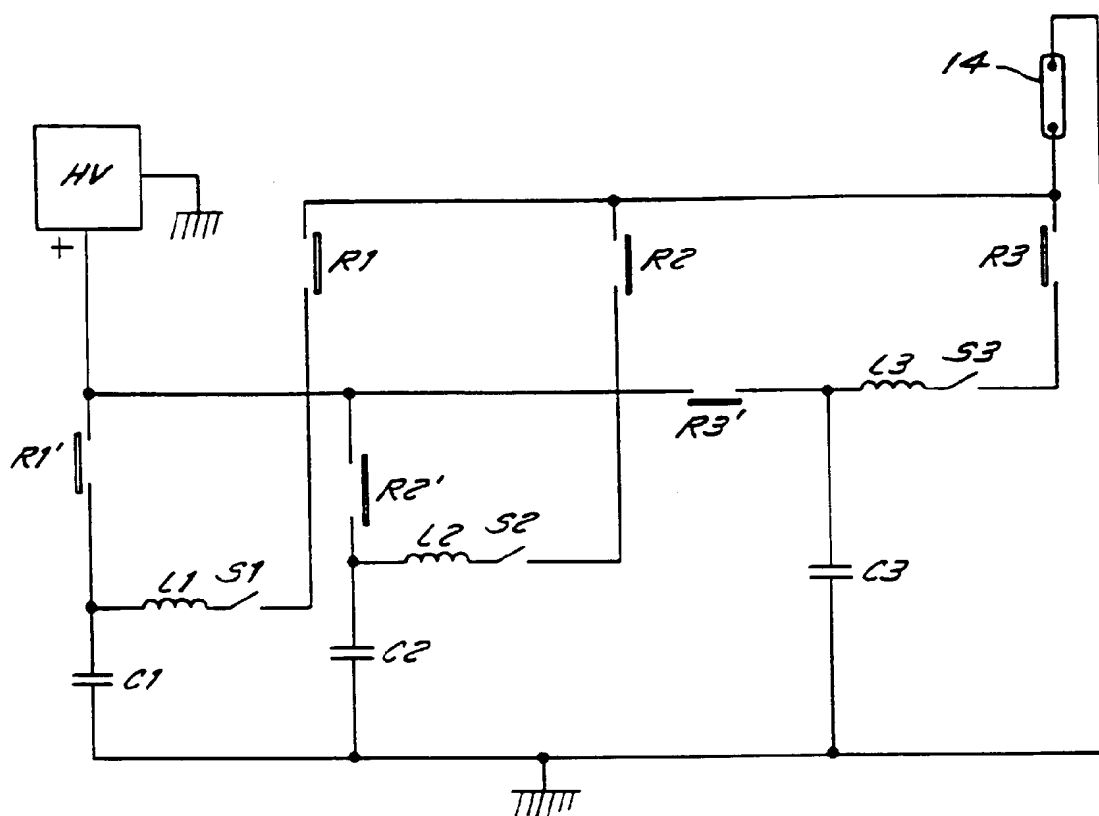


FIG. 3

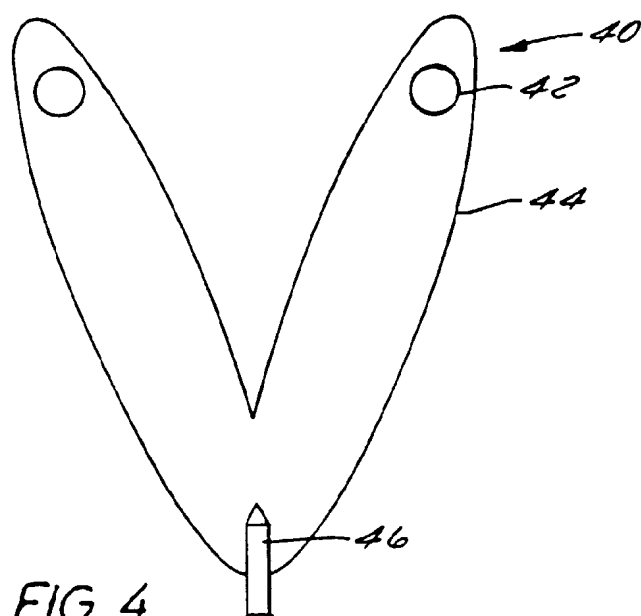


FIG. 4

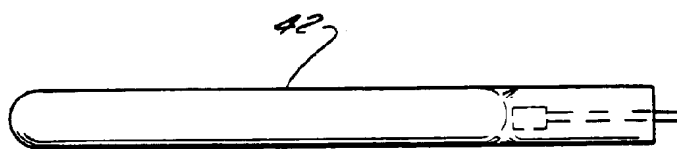


FIG. 5

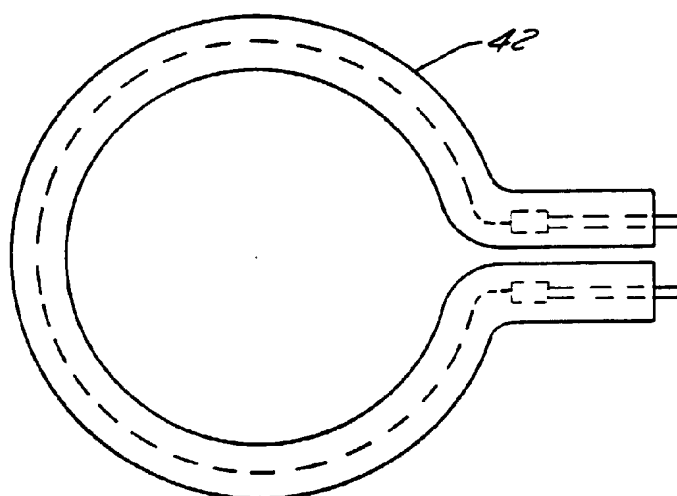


FIG. 6

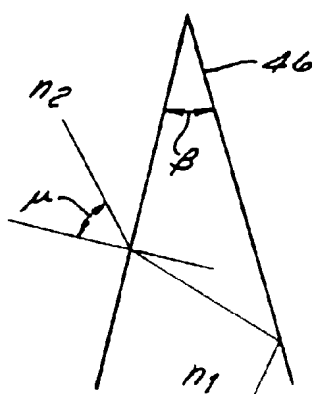


FIG. 7

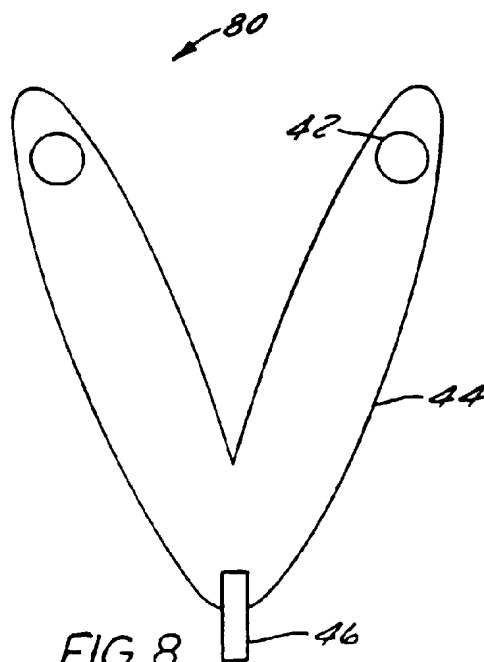


FIG. 8

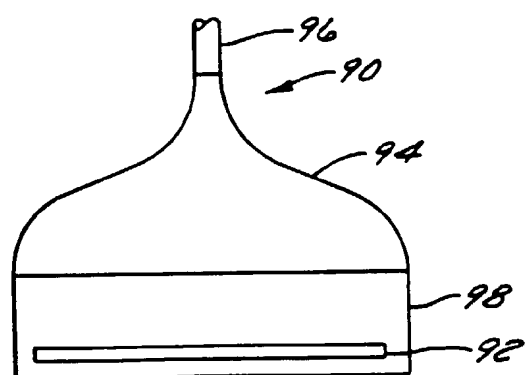


FIG. 9

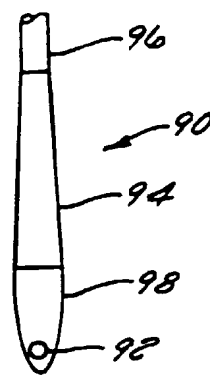


FIG. 10

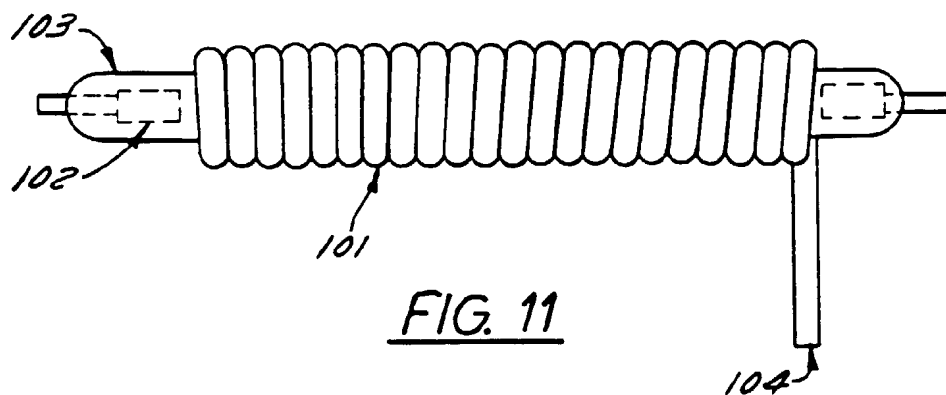


FIG. 11

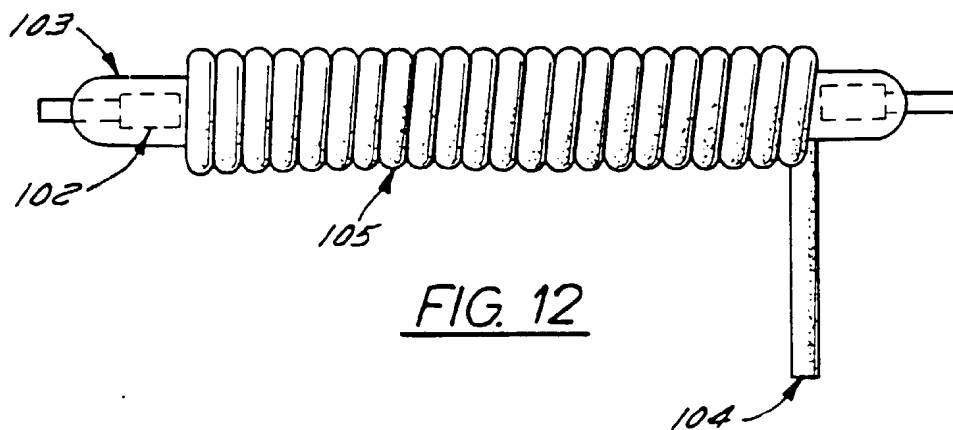


FIG. 12

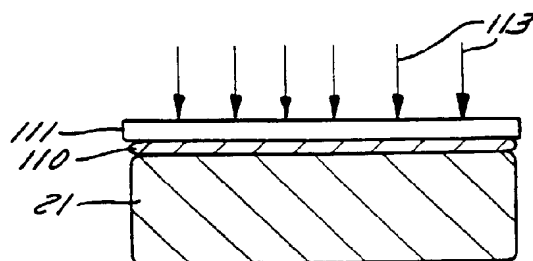


FIG. 13

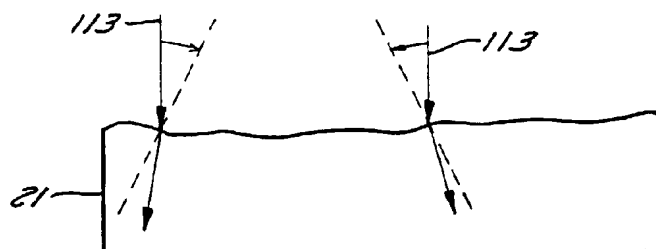


FIG. 14

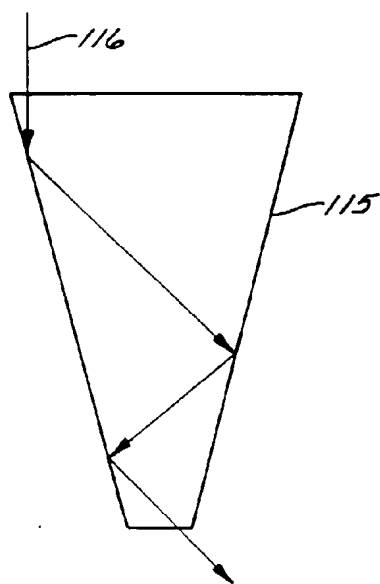


FIG. 15

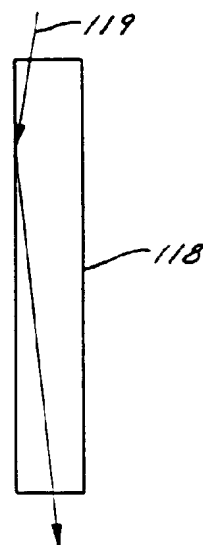


FIG. 16

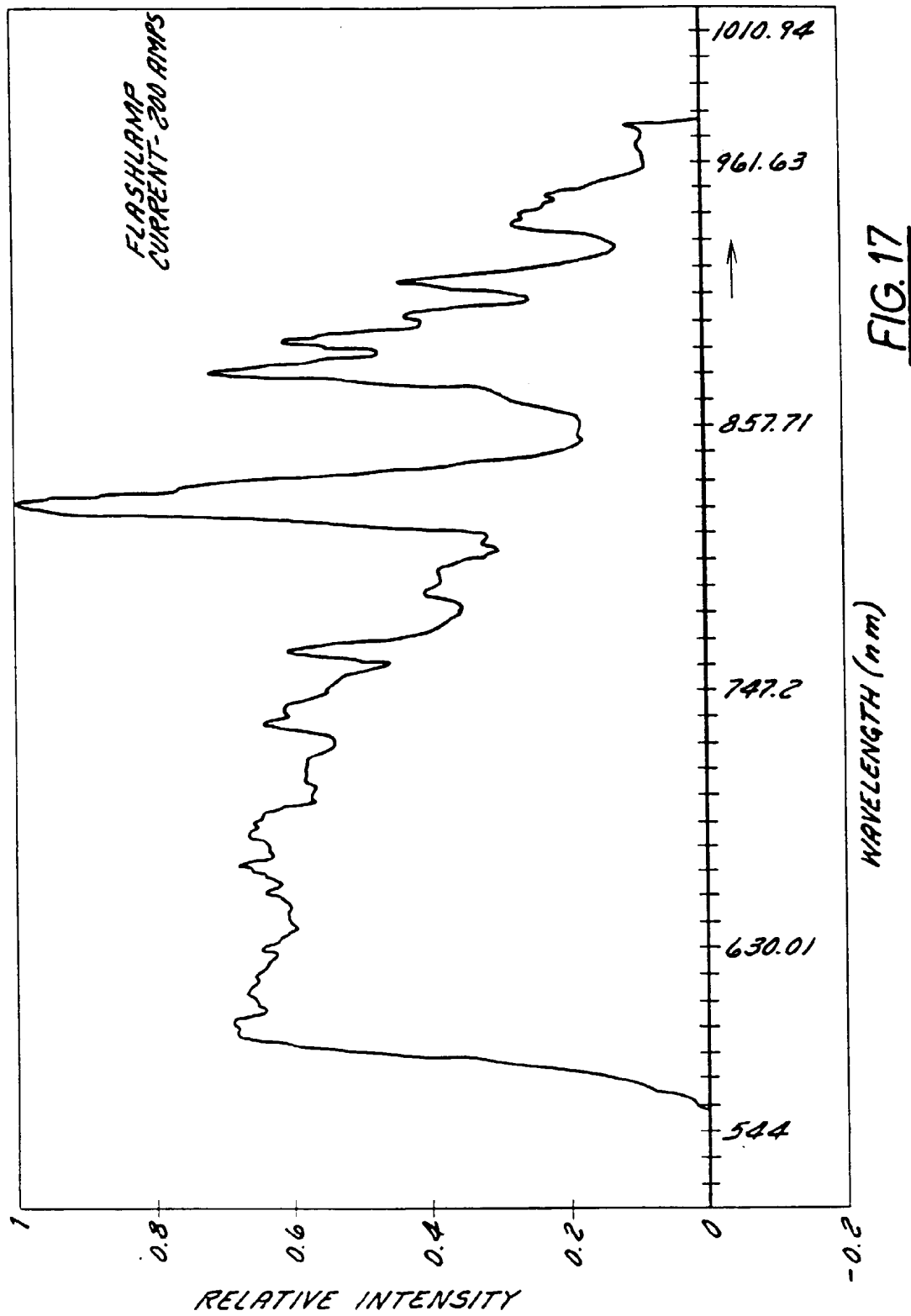


FIG. 17

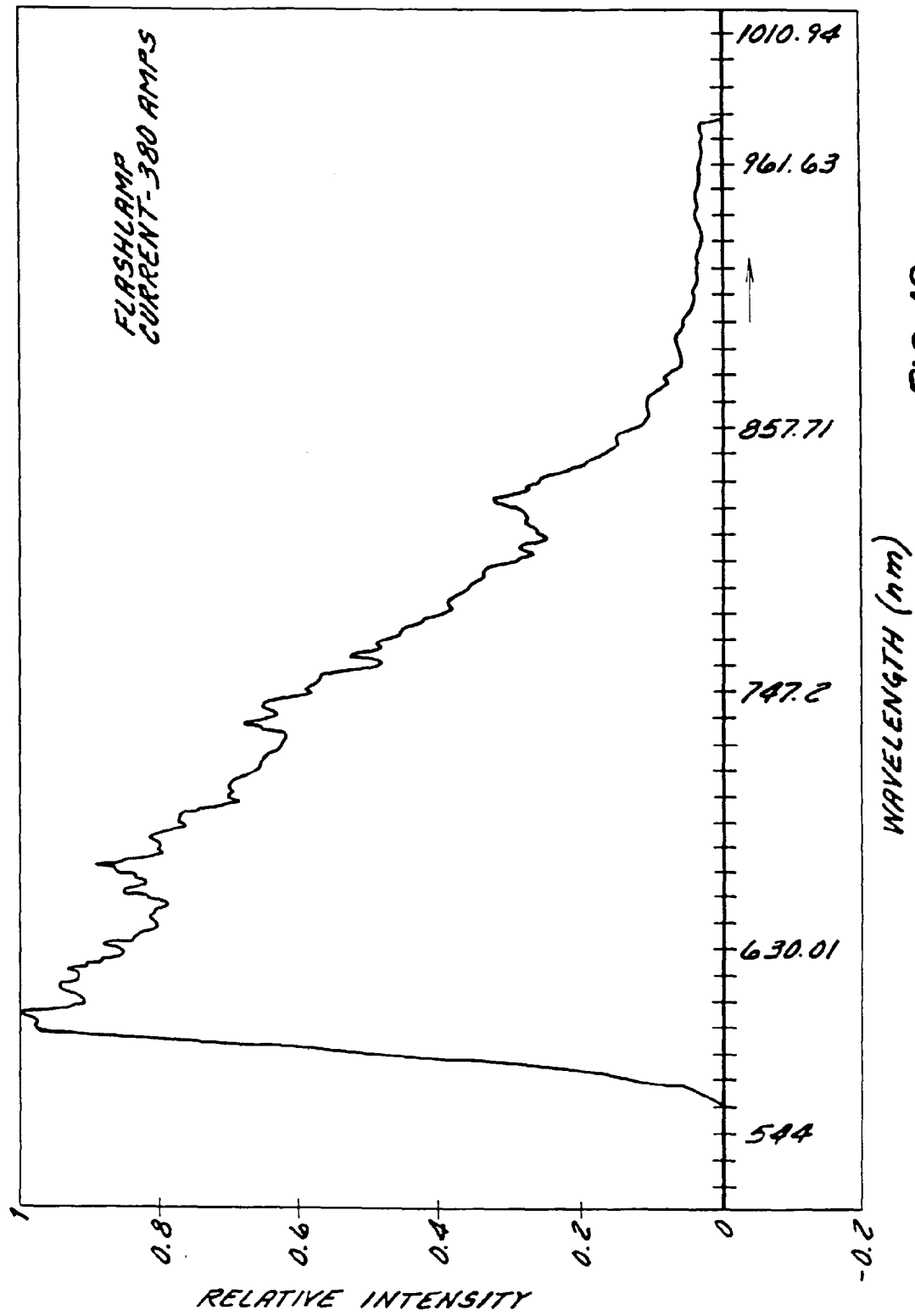


FIG. 18

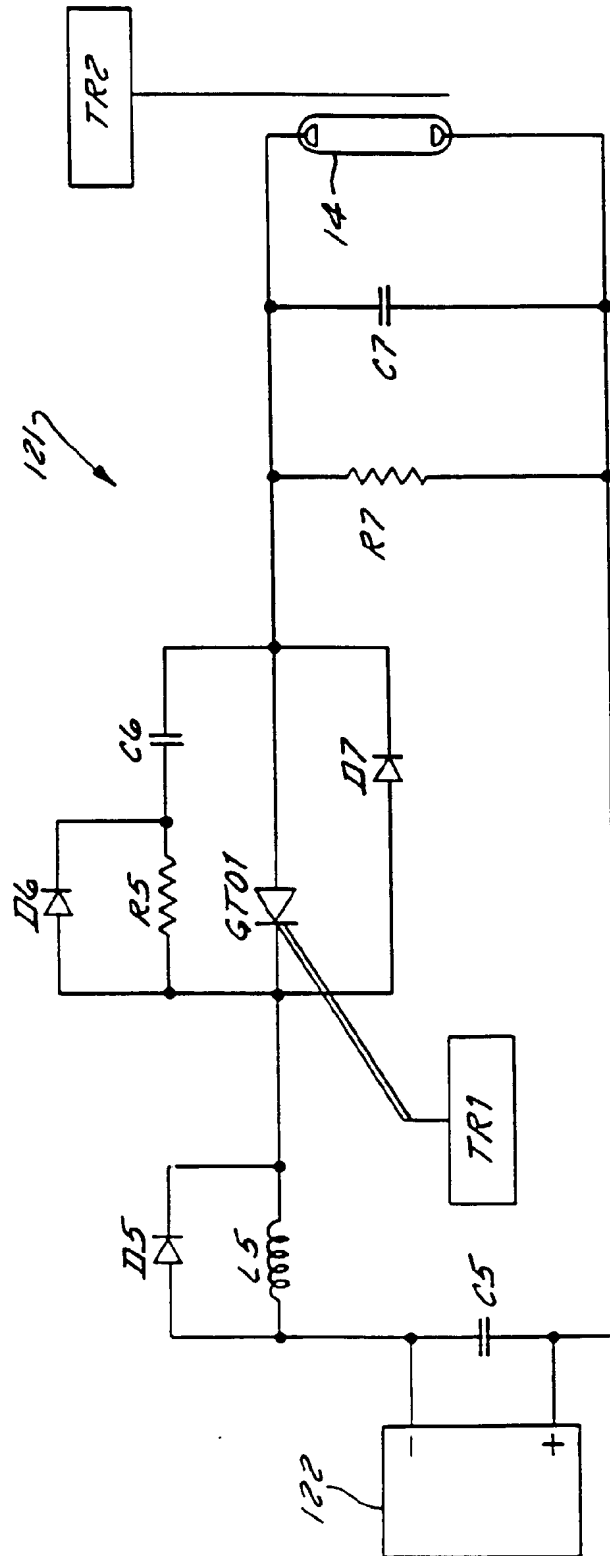


FIG. 19



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(54) **Method and apparatus for skin rejuvenation and wrinkle smoothing**

(57) A method and apparatus for treating skin includes applying pulsed light to the skin to heat and shrinking collagen within the skin, thereby reviving the elasticity of the collagen and of the skin. The epidermis and outer layers of the skin may be protected by cooling with a transparent substance, such as ice or gel, to the skin. The temperature distribution within the skin is controlled by controlling the delay between the time the coolant is applied, and the time the light is applied, by controlling the pulse duration and applying multiple pulses, and by filtering the light and controlling the radiation

spectrum, preferably, the spectrum includes the light having a wavelength in the range of 600-1200nm. The pulsed light may be incoherent, such as that produced by a flashlamp (301), or coherent, such as that produced by a Nd(Yag) laser or a ruby laser, and may be directed to the skin using a flexible or rigid light guide (305).

Also, a method and apparatus for cutaneous resurfacing including directing Er:YAG laser light to the skin. The light may be pulsed, preferably with a delay of about 0.5-10msec between pulses. In one embodiment the pulses have energy fluences of preferably about 100J/cm².

Description

The present invention relates generally to the art of skin treatment using electromagnetic radiation. More particularly, the invention relates to an efficient method and apparatus for skin rejuvenation by ablation of the outer layer of the skin and wrinkle smoothing (or shrinking) by heating of collagen without damage to the epidermis.

There is a strong desire today to obtain and/or maintain a youthful appearance. One manner of doing so is to remove (or reduce) wrinkles. Additionally it is desirable to rejuvenate the skin by removing an outer layer of skin. There are known techniques for removing wrinkles by peeling the skin. Also, there are known methods for rejuvenating the skin. Unfortunately, all known techniques suffer from lack of efficacy and risk to the patient.

One known method of skin rejuvenation includes injection of collagen underneath the skin. This has been performed using a bovine collagen injection. For example, microfine collagen has been injected into periocular lines. Some of the problems with collagen injection include, allergy to collagen and lack of longevity. Also, often there is only partial eradication of the wrinkles.

Peeling most or all of the outer layer of the skin is another known method of rejuvenating the skin. Peeling can be achieved chemically, mechanically or photothermally. Chemical peeling is often carried out using trichloroacetic acid and phenol. An inability to control the depth of the peeling, possible pigmentary change and risk of scarring are among the problems associated with chemical peeling.

The mechanical method is called transcutaneous blepharoplasty and involves shaving off the outer layer of skin. Skin resection during lower lid blepharoplasty frequently results in undesirable side effects, especially ectropion and scleral show. Moreover, transcutaneous blepharoplasty rarely eradicates all of the wrinkle lines.

Pulsed carbon dioxide laser treatment is a known photothermal method of removing of periocular wrinkles. However, laser light is heavily absorbed in water and has a very short range in the epidermis. Thus, a high fluence with short pulse durations will evaporate the outer layer of the skin and peels most or all of the epidermis.

The use of CO₂ laser light for skin rejuvenation also has undesirable side effects. For example, CO₂ lasers have small spot size (3mm or less), and thus their use causes valleys and ridges, particularly when resurfacing large areas. Also, it is difficult to control heat diffusion, and thus the resultant necrosis is difficult to predict and control. Additionally, scar tissue absorbs CO₂ laser light differently than normal skin and thus may adversely impact such a treatment.

Thus, it is apparent there is a need for a new method and device with which it is possible to produce efficient wrinkle removal and skin rejuvenation. This apparatus would preferably be able to control the treatment param-

eters according to characteristics of the tissue, and be easily tunable. The new method and device would preferably provide efficient wrinkle smoothing and skin rejuvenation with minimal side effects.

In accordance with one aspect of the invention a method and apparatus for treating skin includes applying pulsed light to the skin to heat and shrinking collagen within the skin, thereby reviving the elasticity of the collagen and of the skin. In one embodiment the method also includes protecting the epidermis and outer layers of the skin by cooling the epidermis and outer layers of the skin. The cooling may be accomplished by applying a cooled transparent substance, such as ice or gel, to the skin.

In one alternative embodiment the skin is cooled by applying the transparent substance to the skin and then cooling it.

In another alternative embodiment the temperature distribution within the skin is controlled by controlling the delay between the time the coolant is applied, and the time the light is applied. A microprocessor may be used for determining the delay time in response to a selected skin temperature profile. Additionally, the temperature distribution may be controlled by controlling the pulse duration and applying multiple pulses. In another embodiment the temperature distribution within the skin is controlled by filtering the light and controlling the radiation spectrum. Preferably, the spectrum includes light having a wavelength in the range of 600-1200nm.

In another embodiment the pulsed light may be incoherent, such as that produced by a flashlamp, or coherent, such as that produced by an Nd(Yag) laser or a ruby laser.

In another embodiment the light is directed to the skin using a flexible or rigid light guide.

In accordance with a second aspect of the invention a method and apparatus for generating a temperature distribution inside a region of skin having a maximum temperature at a selected depth includes cooling the epidermis and outer layers of the skin and applying pulsed light to the skin.

In one embodiment the cooling is accomplished by applying a cooled transparent substance, such as gel or ice, to the skin. Alternatively, the cooling may be accomplished by applying the transparent substance, and then cooling it.

The temperature distribution is further controlled in one embodiment by controlling the delay between the cooling and the light application. In another embodiment the distribution is controlled by controlling the pulse duration and/or applying multiple pulses.

In accordance with a third aspect of the invention a method and apparatus for cutaneous resurfacing includes directing Er:YAG laser light to the skin. The light may be pulsed, preferably with a delay of about 0.5-10msec between pulses. In one embodiment the pulses have energy fluences of preferably about 100J/cm².

In accordance with a fourth aspect of the invention an apparatus for the cutaneous resurfacing of a region of skin, including skin resurfacing or wrinkle smoothing, includes an incoherent light source such as a flashlamp and an Er:YAG laser. The laser can be operated in a multiple pulse mode. A delivery system delivers the incoherent light and laser light to the region to be treated, and the region may be cooled.

Other principal features and advantages of the invention will become apparent to those skilled in the art upon review of the following drawings, the detailed description and the appended claims.

Figure 1 shows a temperature distribution achieved inside the skin after a cold fluid was applied to the skin, for a plurality of different time delays after the application of the cold gel;

Figure 2 shows a temperature distribution achieved by precooling the skin and applying the light source; Figure 3 is a schematic illustration of the flashlamp light source according to one preferred embodiment of the present invention; and

Figure 4 shows a normalized output filtered radiation spectrum of a flashlamp light source.

Before explaining at least one embodiment of the invention in detail it is to be understood that the invention is not limited in its application to the details of construction and the arrangement of the components set forth in the following description or illustrated in the drawings. The invention is capable of other embodiments or being practiced or carried out in various ways. Also, it is to be understood that the phraseology and terminology employed herein is for the purpose of description and should not be regarded as limiting.

The invention relates to a new method and apparatus of removing wrinkle and rejuvenating skin. Generally, in accordance with this invention, wrinkles are smoothed or reduced by collagen molecules shrinking and increasing the elasticity of the skin and collagen, using a short heating impulse (thermal shock). Tissue is heated at a depth of up to a few millimeters by light radiation, while the skin is externally cooled at the surface to avoid overheating the epidermis. The epidermis may be cooled in a variety of ways, including applying a precooled (i.e., a temperature less than the ambient temperature) transparent substance such as ice or cold gel to the skin. The cold substance should cool the skin before and during treatment. The light (electromagnetic radiation) is applied to the skin in pulses shortly after the application of the cooling material. Alternatively, the fluid or gel could be applied to the skin or skin surface, and then cooled (using thermoelectric cooler, e.g.) shortly before the application of the pulsed light to the skin.

The light source will preferably provide a spectrum such that the optical depth of penetration into the tissue is of the order of 1mm or more. Also, the light source will preferably be able to provide pulses having fluences of

the order of 100J/cm² and peak power of the order of 1000W/cm². A spot size of the order of 10mm is preferable, to reduce scattering losses.

Laser light sources that should be appropriate include a Nd(Yag) laser, a ruby laser, an alexandrite laser, diode lasers and others will be suitable. Incoherent light sources such as a xenon flashlamp should also be appropriate.

A method for cutaneous resurfacing (skin rejuvenation) in accordance with the present invention includes use of an Er:YAG laser light, which has a most efficient wavelength of 2.94μm. Because the absorption depth of an Er:YAG laser in skin is very small (less than 20 microns), it may be difficult to ablate to a depth of the order of 100 microns or more (typical of the epidermis) with it. However, a deeper depth of peeling can be achieved by extending the pulse length of the laser. While this is hard to achieve using an Er:YAG laser due to the inherent short level lifetime, by providing a few pulses with a variable delay between the pulses this limitation may be overcome. Evaporated tissue layer thickness may be controlled by the number of pulses and variation of pulse parameters and delay between pulses.

The invention also relates to an apparatus using a flashlamp light source, or any other source with appropriate parameters, for smoothing wrinkles, without damaging the epidermis. Also, an Er:YAG laser is used for efficient skin rejuvenation by removal of the epidermis.

Generally, the device includes a flashlamp that can provide a pulsed light in the range of 600-1200nm for heating of collagen, a filter system that can cut off the radiation spectrum below approximately 600nm, a light guide that can provide an appropriate spot size and can provide fluences of the order of 100 J/cm², and an Er:YAG laser with pulse energy of the order of 1J, which can be operated in multiple pulse mode with delays between pulses of less than 50msec for skin rejuvenation (by skin ablative peeling).

In one alternative a light source such as a Nd(Yag) laser or ruby laser with appropriate parameters could replace the flashlamp.

This apparatus is very useful for wrinkle removal and skin rejuvenation. A flashlamp light source, particularly when used with external cooling of skin surface, will generate a temperature distribution inside the skin which has a maximum at depth dependent on the light and cooling. Consequently, it is possible to heat collagen molecules without damaging the epidermis. The temperature distribution in the skin is responsive to the delay time between the cooling and application of light, selection of pulse parameters and the radiation spectrum. Accordingly, appropriate control of these parameters allows control of the temperature distribution. An Er:YAG laser operated in multiple pulse mode is very efficient for cutaneous resurfacing procedures and also enables control of depth of evaporation. Thus, the apparatus is safe with little risk of accidental injury to the op-

erator and patient.

As stated above, wrinkles may be smoothed by shrinking collagen molecules using pulsed heating. The present invention method is realized by heating of tissue to depths of up to a few millimeters by light radiation in association with external cooling of skin outer surface to avoid overheating of epidermis. The epidermis may be cooled using many methods. One preferred method is the application of a previously cooled transparent matter like ice or cold gel on the skin which cools the skin before and during treatment. A temperature distribution inside the skin similar to one shown in Figure 1 is created a short time (of the order of 1 second) after the application of the cooled material.

As may be seen, the distribution is such that the epidermis and the outer layer of the skin are colder than the more deeper part of the skin. However, the applied light heats up the superficial parts of the skin more than the inner parts, because of the attenuation of light energy fluence by depth, and due to higher absorption of light by the epidermis.

After heating a temperature distribution such as that shown in Figure 2 results. As may be seen, the deeper parts of the tissue are heated up to a temperature sufficient to cause collagen shrinking, but without damaging the outer parts of the skin (epidermis).

The temperature distribution generated prior to the application of light (Figure 1) is a function of the initial temperature of the cooling material and the delay time between the application of the cooling material and the application of light. By varying this time the depth of penetration of the "cool front" can be varied. When collagen that is deeper needs to be treated without influencing the superficial skin, a longer delay time between the application of the coolant and the light can be used. When the superficial collagen needs to be treated, a shorter delay time can be used.

In a typical treatment the doctor applies the cold gel to the skin before treatment and then applies the light source. In accordance with one embodiment of the invention, the treatment device indicates to the doctor when the light source needs to be applied after application of the cooling material, to achieve a desired temperature distribution. A microprocessor that controls the light generating device may also generate a timing signal for the doctor to accomplish this aspect of the invention.

The applicants have determined that a light source having the following parameters is suitable for implementing the invention.

Light radiation should penetrate into a tissue at a millimeter depth. Examples of light sources which meet the parameter include flashlamp, diode laser, Nd(Yag) laser and ruby laser.

Optical power should be on the order of 100-1000 W/cm².

Fluence should be on the order of 30-150 J/cm².

Spot size should be on the order of a few millimeters

to some centimeters, preferably variable over a range.

A detailed description of one preferred embodiment will be described with reference to Figure 3. As shown in Figure 3, a treatment device 300 includes a flashlamp 301 which can be operated in pulse mode, a reflector 302 which forms a light beam and conducts it to a light guide 305 through a filter system 303 and 304. Reflector 302 is located in a treatment head (or housing) 306.

Filter system 303 and 304 may include one or more filters which cut off the radiation spectrum at approximately 550(or 600)-800nm. Filter 303 provides reflection of the part of unused incident radiation and avoids overheating of absorbing filter 304. Absorbing filter cuts off radiation at approximately 550-800nm. Flexible light guide 305 can be interchanged with a rigid light guide made out of quartz or other types of high optical quality glass. Treatment head 305 is useful for treating large areas.

According to one embodiment, the light energy is applied to the skin using a train of pulses. One advantage of applying a train of pulses is that the epidermis cools relative to the layer of collagen that is heated in the treatment. Preferably, the apparatus produces a train of pulses with variable delays between pulses in the range of 10's to 100's of milliseconds.

The total number of pulses per pulse train can also be varied. More specifically, for a patient with higher skin absorption due to heavier skin pigmentation a larger number of pulses per train is preferably used.

Similarly, the pulse duration of each pulse in the train can also be varied in order to enable cooling of the epidermis without cooling the collagen. In any event, the total dose to the treated area is the product of the number of pulses and the fluence per pulse. The pulse duration, and train length are controlled in one embodiment by a microprocessor 309. As shown on Figure 3, microprocessor 309 provides control signals to pulse forming network 310. Pulse forming network 310 (generally of the type described in commonly owned U.S. Patent No. 5,405,368, which is incorporated herein by reference) provides pulse to flashlamp 301.

The radiation spectrum can be controlled by filter system 303 and 304. Additionally (or alternatively), the spectrum of radiation can be controlled by varying the current density through the flashlamp. If deeper heating is required a longer wavelength radiation is used. Pulse duration may be varied in the range of a few milliseconds to a few ten's of milliseconds.

Other embodiments of the present invention include the use of lasers (those having proper penetration), which can also be very effective to smooth wrinkles. For example, a flashlamp pumped Nd(Yag) laser operating at 1.06μm can provide deep penetration and thus be effective. The laser may be operated in the pulsed train mode, preferably by pulsing the flashlamps that are used to pump the laser. Similarly, a ruby laser may be used. However, the pulse duration cannot be made too long due to the limited value of the lifetime of the lasing

level of these lasers. In the laser embodiment, there is no need for filters since the light is monochromatic. Also this embodiment does not require the use of a rigid light guide since flexible light guides are readily available for laser applications and a low divergence laser beam can be easily focused into a small diameter optical fiber. The use of multiple pulses may be particularly useful to overcome the limited lasing level in the laser embodiment of the invention.

The cutaneous resurfacing method in accordance with the present invention includes an Er:YAG laser light, whose radiation has an absorption depth of much less than that of CO₂ laser radiation, of the order of 50 micron is used. Despite the relatively low absorption depth, an appropriate peeling depth is reached by providing multiple pulses. The thickness of the layer of evaporated tissue may be controlled by the number of pulses, the delay between pulses and varying pulse parameters.

Er:YAG lasers produce radiation of 2.94 μ m, with an energy per pulse of up to 1J. Absorption depth of the radiation is typically about 10 μ m. Thus, to evaporate an epidermis, a train of pulses should be used. Typical delay between the laser pulses should be in the range of 0.5-10msec. The time should preferably be shorter than, or on the order of, the epidermis thermal relaxation time.

Thus, it should be apparent that there has been provided in accordance with the present invention a treatment device that includes a flashlamp or a near infrared pulsed laser in another embodiment, an Er:YAG laser and a coupler that fully satisfy the objectives and advantages set forth above. Although the invention has been described in conjunction with specific embodiments thereof, it is evident that many alternatives, modifications and variations will be apparent to those skilled in the art. Accordingly, it is intended to embrace all such alternatives, modifications and variations that fall within the spirit and broad scope of the appended claims.

Claims

1. An apparatus for treating a region of skin comprising a pulsed light source (301), a housing (306) in which the light source is disposed, wherein the housing (306) includes an aperture suitable for directing the light to the region of skin, and characterised in that the pulsed light source is capable of heating and shrinking collagen in the region of skin, thereby reviving the elasticity of the collagen and of the skin.
2. An apparatus as claimed in claim 1 further including a timing circuit coupled to the pulsed light source, adapted to indicate when a delay time has passed after an application of a cooling substance to the skin region.
3. An apparatus as claimed in claim 1 or 2 further comprising a microprocessor (309) coupled to the pulsed light source for determining the delay time in response to a selected skin temperature profile.
4. An apparatus as claimed in claim 2 or claim 3 when appended to claim 2 further comprising a means for reducing the temperature of the cooling substance, wherein the cooling means is disposed to provide a signal indicative of cooling to the timing circuit.
5. An apparatus as claimed in any one of claims 1 to 4 further comprising a pulse formation circuit and a pulse duration input, wherein the pulse duration circuit is coupled to the pulsed light source and is adapted to control the duration of pulses emitted by the pulsed light source in response to the pulse duration input.
6. An apparatus as claimed in any one of claims 1 to 5 wherein the pulsed light source includes a noncoherent light source.
7. An apparatus as claimed in any one of claims 1 to 6 further including a filter (303, 304) disposed adjacent to the aperture, wherein a temperature distribution within the skin is controlled in response to a radiation spectrum produced by filtering the light.
8. An apparatus as claimed in any one of claims 1 to 7 further including a light guide (305) disposed adjacent to the aperture.
9. An apparatus as claimed in any one of claims 5 to 8 further including a pulse delay circuit adapted to produce a delay in the range of 0.5-10msec between successive pulses of light emitted by the pulsed light source pulses.
10. An apparatus as claimed in any one of claims 1 to 9 wherein the light source is adapted to provide pulses having energy fluences on the order of 100J/cm².
11. A cosmetic treatment of a region of skin comprising the steps of applying pulsed light, heating collagen and shrinking collagen, thereby reviving the elasticity of the collagen and of the skin.
12. A cosmetic treatment as claimed in claim 11 further comprising the step of protecting the epidermis and outer layers of the skin by cooling the epidermis and outer layers of the skin.
13. A cosmetic treatment as claimed in claim 12 in which the step of cooling includes the step of applying a transparent substance having a temperature less than an ambient temperature, to the region of

the skin.

14. A cosmetic treatment as claimed in claim 13 further including the step of controlling a delay time between the application of the substance and the application of light, to control the temperature distribution within the skin. 5
15. A cosmetic treatment as claimed in any one of claims 11 to 14 further comprising the steps of controlling a pulse duration and applying multiple pulses to control a temperature distribution within the skin. 10
16. A cosmetic treatment as claimed in any one of claims 11 to 15 wherein the step of applying pulsed light includes the step of pulsing a noncoherent light source. 15
17. A cosmetic treatment as claimed in any one of claims 11 to 16 further including the step of controlling the radiation spectrum by filtering the light to control a temperature distribution within the skin. 20

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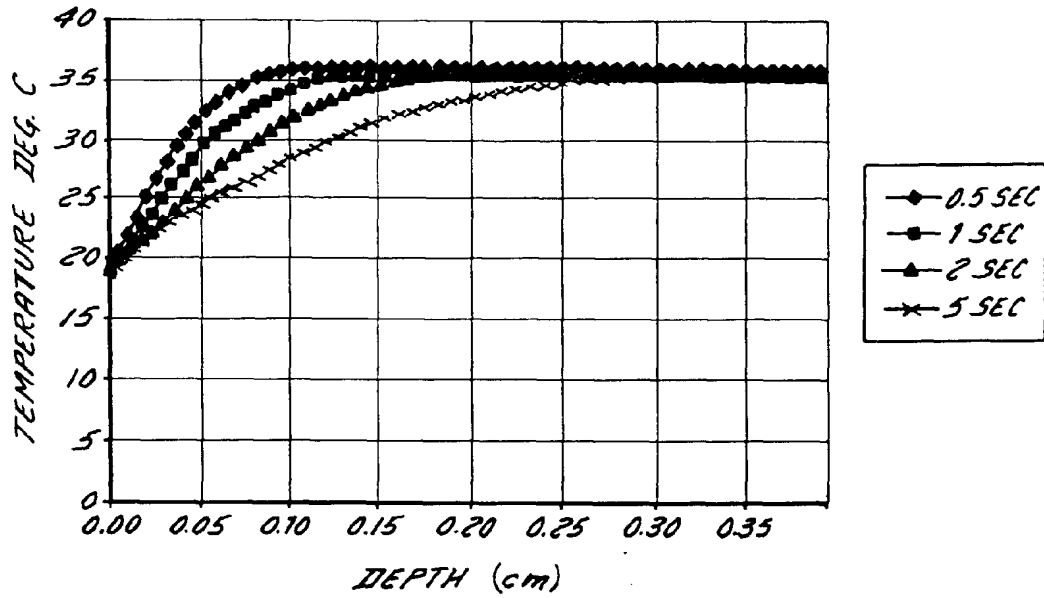


FIG. 1

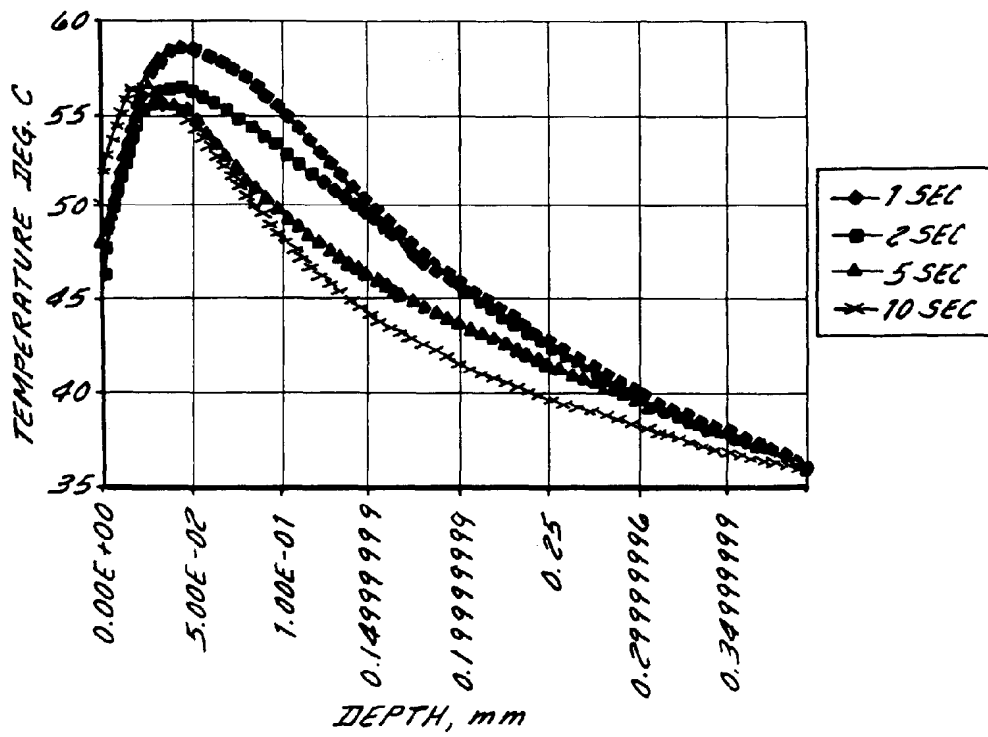


FIG. 2

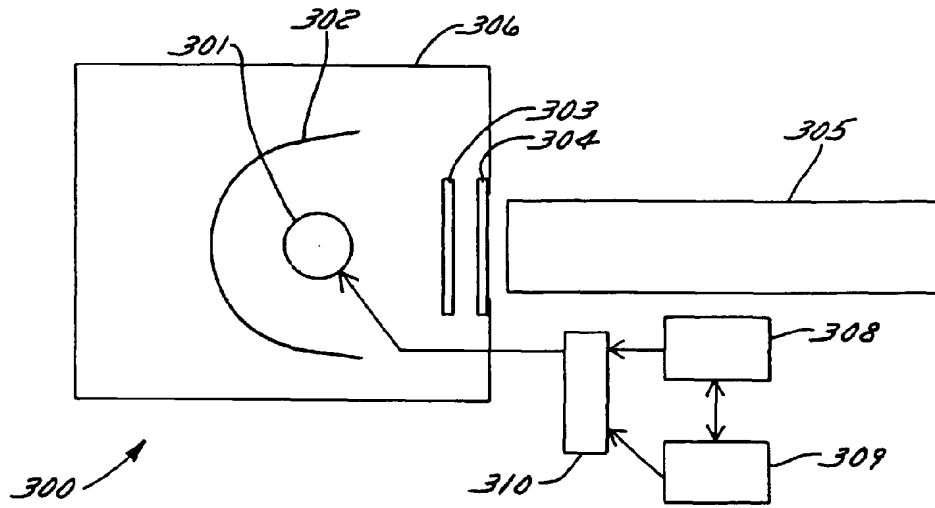


FIG. 3

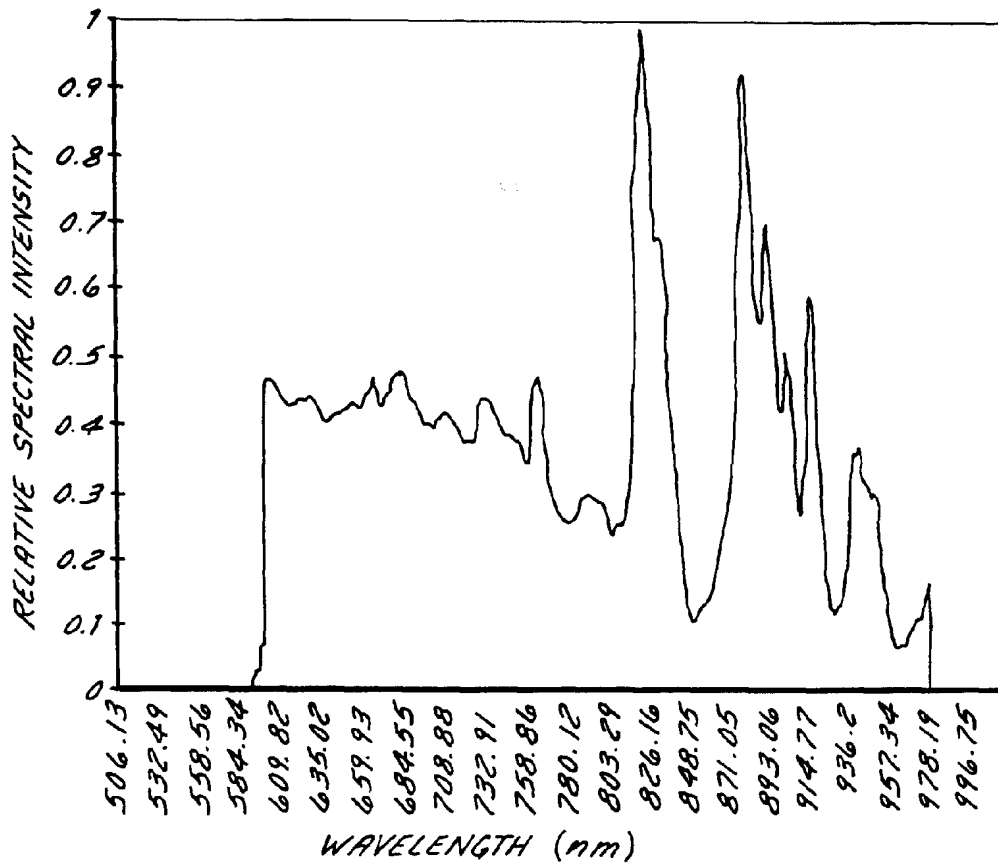


FIG. 4



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(54) **Method and apparatus for treatment of cancer using pulsed electromagnetic radiation**

(57) The invention includes a method for the hyperthermic treatment of tumors including the steps of providing a pulsed radiation output from a radiation source; and directing said pulsed radiation output toward a tumor. The invention further includes an apparatus for use in the treatment of tumors having a radiation source (14) adapted to produce broad-band pulsed radiation output at least in the visible and near-infrared range of wave-

lengths, a delivery system proximal to the radiation source and adapted to focus and direct the pulsed radiation output to a dermal treatment site, and a filtering system adapted to restrict the pulsed radiation output to bands in the visible and near-infrared range of wavelengths. In particular the radiation source is adapted to produce pulsed radiation output over a continuous band of wavelengths between 600 nm and 1000 nm.

Description

This invention relates to an apparatus and method for the treatment of tumors. More particularly, the invention relates to an apparatus for the irradiation of shallow tumors with pulsed electromagnetic radiation.

Several non-surgical methods are available for treatment of cancer, but all of them have disadvantages. Chemical therapy and photodynamic therapy are accompanied by the introduction of a toxic agent into the body. Electromagnetic radiation therapy using X-rays causes the destruction of healthy tissue due to X-rays ability to penetrate deeply into human tissue.

Another method, called hyperthermia, is used for tumor necrosis both by itself, and in combination with other methods of cancer treatment. The basic purpose of hyperthermia is to raise tumor temperature substantially above body normal temperature, to a temperature at which tumor cells are killed. The "selectivity" of hyperthermic therapeutic methods are the extent to which the tumors and not the surrounding healthy tissue is destroyed. Hyperthermic treatments have been employed for both whole body heating and for local heating of tumors. Local hyperthermia typically uses sources of electromagnetic radiation, focused on the tumor at frequencies that will heat tumor tissue and not the surrounding healthy tissue. Microwave, visible and infrared frequency ranges are commonly employed for this purpose.

Current hyperthermic methods have significant disadvantages. Treatment times are often long, on the order of an hour. Furthermore, the selectivity of the radiation is low, causing necrosis not only of tumor tissue, but of the healthy surrounding tissue as well.

Hyperthermia treatments using microwave radiation sources (typically radiating at about 915 MHz) have the disadvantage of deep non-tunable penetration (several centimeters) into the body as well as problems with focusing which cause low selectivity.

Nd:YAG laser radiation sources are used both by themselves and in combination with photodynamic therapy. One disadvantage of Nd:YAG laser when used for hyperthermia is its small spot size, on the order of 5 mm. A radiation source this small cannot easily heat large tumors, which may have a projected area of several square centimeters on the skin, resulting in extended treatment times. In addition, the Nd:YAG laser has other limitations relating to their continuous wave (CW) operating mode, and with their limited tunable range. It is clear that an improved apparatus and method for hyperthermia tumor treatment is desirable.

Pulsed radiation of a tumor using a light source would cause more efficient hyperthermia and necrosis than current methods provide. Furthermore, a radiation source capable of heating tissue in a short time interval, preferably between 41 and 45 degrees C, would reduce the treatment times currently required. Providing a radiation source with a broad controllable spectrum of radiation in the visible and near infrared regions would allow

the penetration depth and the selectivity of the treatment to be more accurately controlled.

The present invention is directed to a method for the hyperthermic treatment of tumors with electromagnetic radiation including the steps of providing a pulsed radiation output from a radiation source and directing said pulsed radiation output toward a tumor. The radiation may be developed over at least one continuous band of wavelengths, or be generated in the visible and near-infrared band, possibly in a continuous band between 600 and 1000 nm. In one embodiment, it may include the step of transmitting a broad radiation beam to a pigmented tumor, which might have a cross-sectional area of between 0.8 cm² and 500 cm². In another embodiment, it is possible to control the pulse-width of the pulsed radiation output, focus the radiation source for controlling the power density of the pulsed radiation output, or filter and control the spectrum of the pulsed radiation output. In particular, one may focus the pulsed radiation output to a beam having a cross-sectional area of greater than 0.8 cm². Alternatively, one may cut off the UV portion of the spectrum. A pulse width in the range of about 100 microseconds to 50 milliseconds may be provided, particularly, one having an energy density at the treatment area of at least 0.2 W/cm². Alternatively, energy densities of greater than 90 J/cm², 120 J/cm² per treatment may be provided at the treatment site. A pulse delay of greater than 100 milliseconds or less than 100 seconds may also be provided.

In another embodiment of the invention, an apparatus for the treatment of tumors is provided, including a radiation source producing pulsed radiation at least in the visible and near-infrared wavelengths, a delivery system near the radiation source for focusing and directing the radiation to a treatment site, and a filtering system restricting the radiation to visible and near-infrared wavelengths. Alternatively, the radiation source may produce pulsed radiation in a broad band, or over at least one continuous range of wavelengths. This may be focused in a beam of at least 0.8 cm². The radiation may be restricted to a band between 300 and 1000 nm, or may be UV blocked by a filter. The radiation pulses may have a duration of between 100 μsecs and 100 msecs, and may be spaced from 100 msecs to 100 secs apart. In addition, they may be delivered to the treatment area with a radiation density of greater than 0.2 W/cm², 90 J/cm², or 120 J/cm². The radiation may also be limited to a radiation density of less than 200 J/cm².

Other principal features and advantages of the invention will become apparent to those skilled in the art upon review of the following drawings, the detailed description and the appended claims.

The present invention will now be described, by way of example only, with reference to the accompanying drawings, in which:-

FIGURE 1 is a graph of radiation tissue penetration versus radiation wavelength;

FIGURE 2 is a cross-sectional view of tumor treatment device according to the present invention; and **FIGURE 3** is a graph of treatment results using the **FIGURE 2** tumor treatment device.

Before explaining at least one embodiment of the invention in detail it is to be understood that the invention is not limited in its application to the details of construction and the arrangement of the components set forth in the following description or illustrated in the drawings. The invention is capable of other embodiments or being practiced or carried out in various ways. Also, it is to be understood that the phraseology and terminology employed herein is for the purpose of description and should not be regarded as limiting.

The present invention is directed to a method and apparatus for treating shallow tumors using pulsed radiation. Treatment of such tumors is problematic, since the outer layers of skin must be penetrated and not harmed, yet the radiation must get to the underlying tumorous growth sufficient to heat the tumor and cause necrosis. The "effective penetration depth", **d**, of radiation is a measure of the radiation's ability to penetrate the skin and affect an underlying tumor. It is defined herein as the depth below the surface of the skin at which the radiation fluence reaches **1/e** times the magnitude of the radiation fluence on the surface of the skin. Since the effective penetration depth varies with the wavelength of the impinging radiation, tumors at a particular depth can be targeted, and the overlying skin preserved, by selecting and applying particular wavelengths of radiation for tumors at a particular depth.

The effective penetration depth can be estimated by using the effective attenuation coefficient, μ_{eff} , of the dermis, which takes into account the scattering and absorption of light in tissue. The relation of the effective penetration depth to the effective attenuation coefficient can be estimated as:

$$d = 1/\mu_{\text{eff}}$$

Following Jacques (S. L. Jacques, Role of Skin Optics in Diagnostic and Therapeutic Uses of Lasers, "Lasers and Dermatology", Springer-Verlag, 1991, pp. 1-21), the effective attenuation coefficient of the dermis can be expressed as follows:

$$\mu_{\text{eff}} = \{3 \mu_a (\mu_a + \mu_s (1-g))\}^{1/2},$$

where

μ_{eff} = attenuation coefficient of dermis
 μ_a = absorption coefficient of dermis
 μ_s = scattering coefficient of dermis, and
 g = the anisotropy factor, defined as the average cosine of the scattering angle for one scatter-

ing event.

Using the above coefficients and factor, a chart has been made of the effective penetration depth in centimeters versus the wavelength of electromagnetic radiation impinging upon the skin. This chart is illustrated in **FIGURE 1**. As **FIGURE 1** discloses, the effective penetration depth increases with increasing wavelength, and for wavelengths between 400 nm and 1000 nm varies between 0.03 cm and 0.25 cm. Radiation can penetrate as deeply as 2 mm with a radiation wavelength of 800 nm. The sensitivity of effective penetration depth to wavelength is clear from this chart. For example, **d** doubles when the wavelength of the impinging radiation increases by a mere 20% (500 to 600 nm). Because varying the applied radiation wavelength varies the depth of penetration of that radiation, one can control treatment depth by controlling the radiation wavelength.

Hyperthermic treatments also depend upon the length of time radiation is applied to the surface of the skin. The effective depth of tissue heating based on heat conducted from the surface depends upon the conductivity of the skin. The time **t**, required for a heat wave to penetrate to a depth **d**, below the surface of the skin can be expressed as:

$$t = d^2/a,$$

where:

a = the diffusivity of the skin (approximately $3 \times 10^{-7} \text{ m}^2 \text{ sec}^{-1}$).

Thus, the depth of penetration can be controlled by controlling the time interval over which radiation is applied to the surface of the skin. For example, conducting heat from the surface of a skin throughout a shallow tumor with a thickness of about 1 cm requires about a 5 minute application of radiation to the surface of the skin.

These two modes of heating: conduction from the surface of the skin, and radiant penetration, can be tailored to specific tumors by varying the wavelength and the pulse duration.

A major limitation to the use of radiation sources for therapeutic treatment is the potential tissue damage. In order to radiate the tumor with the optimum wavelengths of radiation yet not burn tissue, a radiation source is preferably pulsed, thereby providing radiation at wavelengths sufficient to penetrate the tumor to an optimum depth, yet limiting the average energy density during a treatment and preventing the upper layers of the tumor from being overheated.

To provide for the treatment of a wide range of shallow tumors, the preferred energy density per pulse is between 0.1 and 10 Joules per square centimeter of tumor area. These pulses are preferably repeated at a rate of between 0.1 and 1 Hertz. The number of pulses for

treating shallow tumors preferably ranges between 1 and 1000 pulses. To treat a wide range of tumor sizes, the radiation should be applied to an area of the skin ranging from 0.8 cm² to 500 cm².

It is clear from FIGURE 1 that by irradiating a tumor with selected bands of radiation in the visible and near infrared regions, the tumor can be penetrated to a depth of between 0.05 and 0.25 cm and hyperthermally treated. FIGURE 2 illustrates just such a tumor treatment apparatus 10, having a housing 12 that encloses a radiation source 14, and a reflector 16, and having an opening with a set of optical filters 18, 20, and a delivery system 22. A processor 24 is provided to control radiation source 14 through lamp driver circuit 26, under the control of a program in memory 28.

Radiation source 14 is a flashlamp such as a gas filled linear flashlamp Model No. L5568 available from ILC. Typically, a flashlamp's energy is emitted as broad-band incoherent energy in the 300 to 1000 nm wavelength range, which, as FIGURE 1 shows, is well-suited to penetrating tissue to a depth of several millimeters, and thus, for treating shallow tumors.

To treat a tumor, the radiation must be focused and delivered to the treatment site, and thus reflector 16 and delivery system 22 are provided. Reflector 16 gathers the radiation and directs it toward an opening in the housing. To effectively reflect radiation in the 300 to 1000 nm band, reflector 16 is preferably metallic, typically aluminum which is easily machinable and polishable, and has a very high reflectivity in the visible and near infrared ranges of the spectrum. Other bare or coated metals can also be used for this purpose.

Optical filters 18 and neutral density filters 20 are mounted in housing 12 and may be moved into the beam or out of the beam to control the spectrum and intensity of the light. The optical filters may include bandwidth and low cutoff filters in the visible and infrared portions of the spectrum. To limit skin damage, it is desirable to employ UV filters to block the UV portion of the spectrum, in particular, UV filters that cut off the spectral range below 510 nm. For deeper penetration it is preferable to use narrower bandwidth filters. Optical bandwidth filters and the cutoff filters are readily available commercially. Neutral density filters with varying degrees of filtration can be used to reduce the total fluence transmitted to the skin by blocking the transmission of radiation emitted by the radiation source to the treatment site.

The radiation is delivered to the treatment site by delivery system 22, typically an optical fiber or a quartz light guide, although it may be preferable to emit light directly from an opening in the housing. The delivery system should produce fluences on the skin of between 100 mJ/cm² to 10 J/cm².

Radiation source 14 is pulsed to provide control of the total fluence, and thus control of tumor and skin heating. To vary the fluence, the delay interval between pulses may be increased or decreased, preferably over a range of a hundred milliseconds to tens of seconds. In

this manner, the tumor can be heated at a rate sufficient to allow skin penetration and tumor necrosis, yet not overheat tissue. Total fluence can also be controlled by varying the duration of each pulse over a range of between a hundred microseconds and tens of milliseconds, to vary the fluence per pulse from a hundred millijoules to tens of Joules using a flashtube. Total fluence can also be modified by varying the energy per pulse.

Effective penetration depth is dependent on the wavelength of radiation received at the surface of the skin. The present invention provides for changes in wavelength in several ways. Filter 18 can be a low-pass or band-pass filter, thereby blocking selected wavelengths of light. Varying the power per pulse will also vary the emission spectrum of the radiation source as well.

Processor 24 is provided to control the energy per pulse, the pulse repetition rate, pulse duration rate and the number of pulses per a single treatment. It is connected to radiation source 14 through a lamp driver circuit 26, which is capable of generating power sufficient to trigger radiation source 14. Processor 24 operates under the control of a program stored in memory circuit 28.

The present invention is well suited to treating tumors with a wide variety of sizes. For smaller tumors, a fiber optic delivery system is appropriate. By directing the radiation through a fiber to the treatment site, small tumors typically on the order of a millimeter or larger in breadth can be treated without endangering the surrounding tissue. Larger tumors, typically on the order of several square centimeters in projected area, can be treated using a delivery system, that focuses and applies the radiation to a wider treatment site, preferably radiating a 0.8 cm² area of the treatment site or larger. By applying the radiation over a larger area, for example 500 cm², even heating of large tumors can be achieved, reducing the chance of uneven tumor treatment and the risk of damaging tissue.

The present invention has been tested in animal trials and is effective for the treatment of tumors. FIGURE 3 illustrates the inhibition of melanoma B16 growth in mice after irradiation in accordance with this invention. The FIGURE 3 chart compares tumor volume versus time for three irradiation levels: a control level (0 J/cm²); 90 J/cm²; and 120 J/cm². Irradiation levels of 90 J/cm² clearly and significantly delay tumor growth, and an irradiation level of 120 J/cm² causes the affected tumor to shrink in size. Extrapolating from these tests, irradiation levels of 200 J/cm² are believed to provide therapeutic results. The tumor treatment apparatus in these tests applied broad-band radiation in the band from 600 nm to 1000 nm to the tumor. No apparent tumor response was observed for average radiation power densities below 0.2 W/cm².

Thus, it should be apparent that there has been provided in accordance with the present invention a method and apparatus for the hyperthermic treatment of tumors

that fully satisfies the objectives and advantages set forth above. Although the invention has been described in conjunction with specific embodiments thereof, it is evident that many alternatives, modifications and variations will be apparent to those skilled in the art. Accordingly, it is intended to embrace all such alternatives, modifications and variations that fall within the spirit and broad scope of the appended claims.

and 50 seconds.

10. The use of apparatus according to any one of claims 1 to 9 in a method of treatment of tumors.

Claims

1. An apparatus for the hyperthermic treatment of tumors comprising: a radiation source (14) adapted to produce pulsed radiation output over a continuous band of wavelengths between 600 nm and 1000 nm at least in the visible and near-infrared at an intensity sufficient to cause tumor necrosis; and a delivery system proximal to the radiation source and adapted to direct the pulsed radiation output to a dermal treatment site.
2. An apparatus as claimed in claim 1, further comprising a filtering system adapted to restrict the pulsed radiation output to bands in the visible and near-infrared range of wavelengths.
3. An apparatus as claimed in claims 1 or 2 wherein the delivery system is adapted to direct the pulsed radiation output to a beam having a cross-sectional area at a treatment site of at least 0.8cm².
4. An apparatus as claimed in claims 2 or 3 wherein the filtering system includes a filter adapted to block UV wavelengths.
5. An apparatus as claimed in any one of claims 1 to 4 wherein the delivery system is adapted to deliver the pulsed radiation output to the treatment area with a radiation density of greater than 0.2 W/cm².
6. An apparatus as claimed in any one of claims 1 to 5 wherein the delivery system is adapted to deliver the pulsed radiation output to the treatment site with a radiation density of greater than 90 J/cm².
7. An apparatus as claimed in any one of claims 1 to 5 wherein the delivery system is adapted to deliver pulsed radiation output to the treatment site with a radiation density of greater than 120 J/cm².
8. An apparatus as claimed in any one of claims 1 to 7 further including a processor adapted to control the pulse duration and pulse delay.
9. An apparatus as claimed in any one of claims 1 to 8 wherein the pulsed radiation source is adapted to provide a pulse delay of between 100 milliseconds

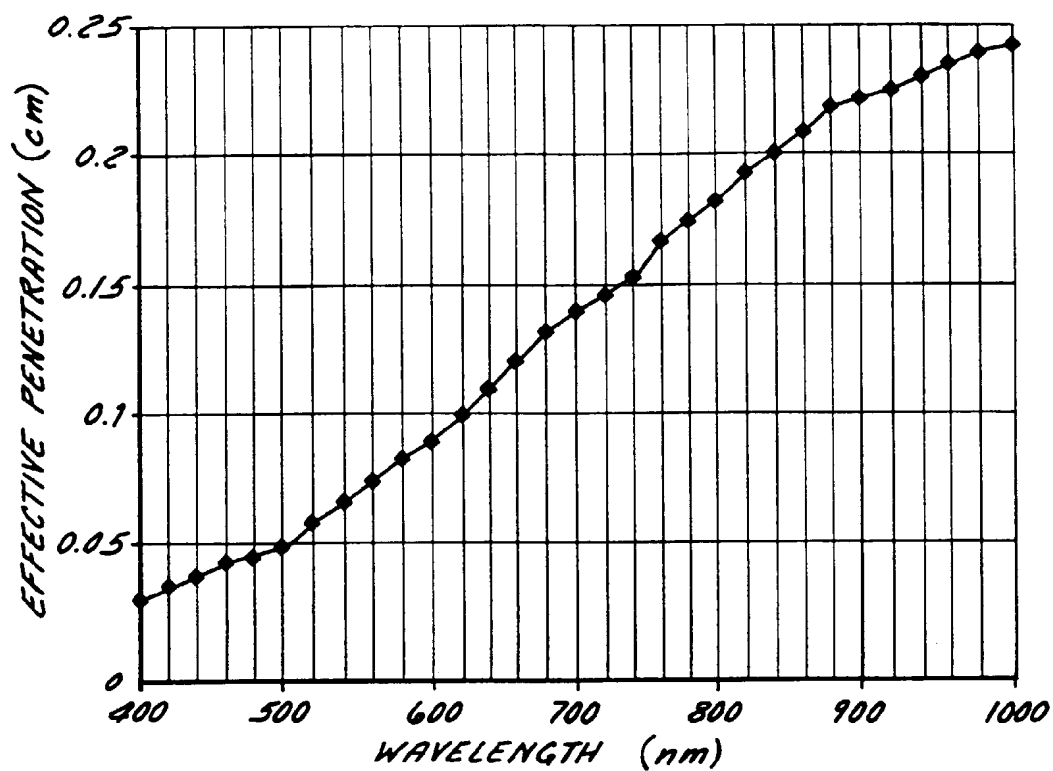


FIG. 1

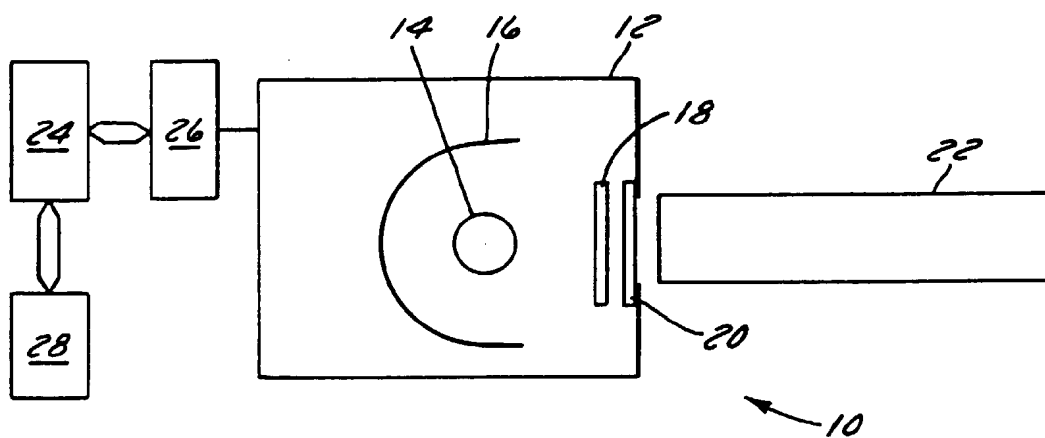
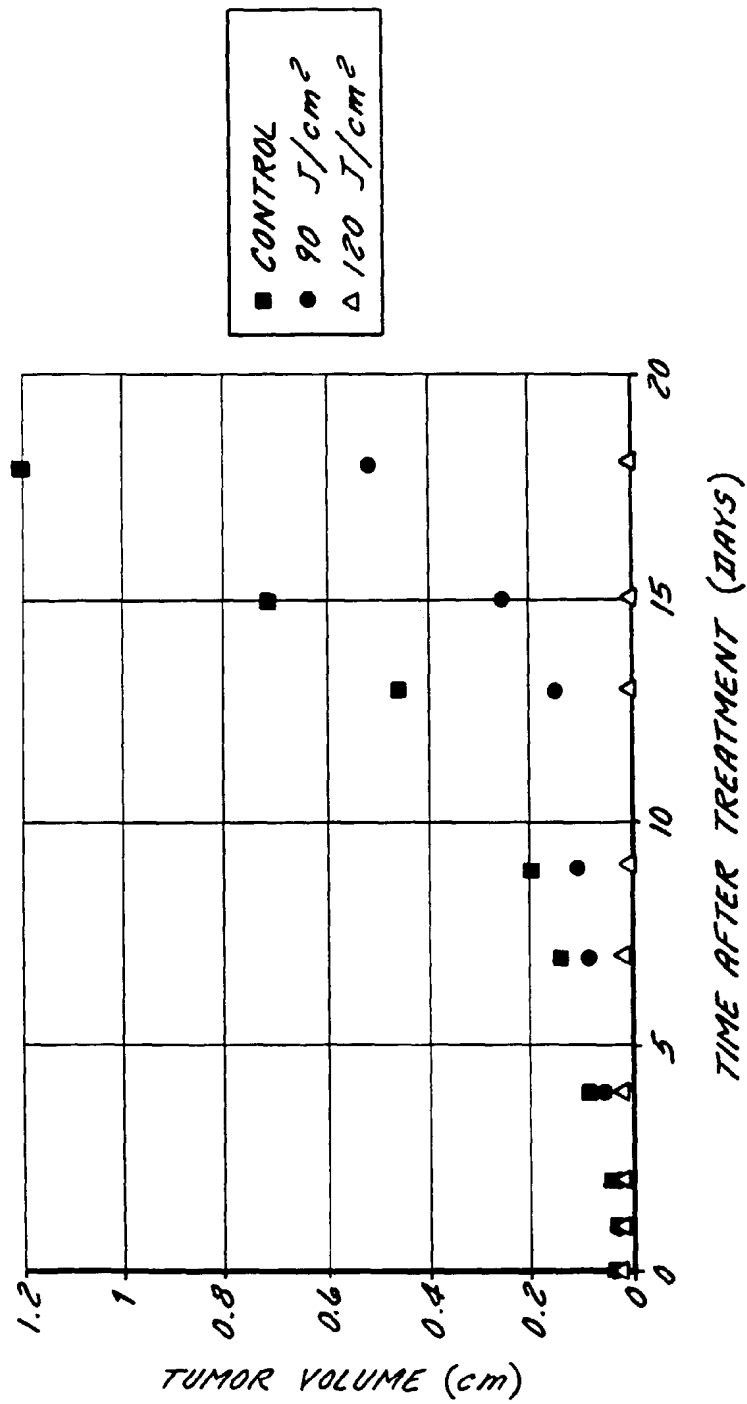


FIG. 2

FIG. 3



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(54) **Method and apparatus for treating psoriasis using pulsed electromagnetic radiation**

(57) A method and apparatus for treating psoriasis includes a source of incoherent electromagnetic energy. The energy is directed to a region of tissue to be treated. The pulse duration and the number of pulses may be selected to control treatment parameters such as the heating of healthy tissue and the penetration depth of the energy to optimise the treatment. Also, the radiation may be filtered to control the radiation spectrum and penetration depth. The filtering may include attenuating an UV portion of the radiation spectrum and portions of the spectrum below a desired treatment bandwidth. A light guide for large or small spot sizes may be used to

direct the light to the skin. A cooling gel is applied to the skin to be treated in another embodiment. The gel may be cooled either before or after it is applied to the skin.

The use of apparatus comprising a housing (5), including a window (11), a light source (1) disposed within the housing, wherein the light source is capable of producing incoherent radiation in the visible and near infra-red range of wavelength, a reflector (2) disposed within the housing and further disposed to direct the light through the window, and a filtering system disposed in the path of the radiation in a method of treatment of psoriatic plaque.

Description

The present invention relates generally to a method and apparatus for treating psoriasis. More particularly, the invention relates to treating psoriasis by irradiating psoriatic plaques with visible and near infrared electromagnetic radiation.

Psoriasis is a relatively common skin disease that appears in a few percent of the population. Prior art treatments of psoriatic plaques fall generally into two categories: the use of a topical drug (i.e., a drug that is applied to the skin externally) and the application of light to the psoriatic plaques. However, many of these prior art techniques have a common fundamental drawback: they offer relief to the patient for only a limited time.

The topical drug treatment includes treatment by coal and wood tar dithranol and corticosteroids. These treatments produce acceptable results that last for no more than a few weeks. Detergent shampoos, salicylic acid ointments are also used, but they are also limited in their efficiency and the length of time of relief for the patient.

Treatments utilizing light generally use a portion of the electromagnetic spectrum. For example, ultraviolet light in the UVA and UVB ranges is used extensively. This method of treatment offers a very limited relief to the patient and is used on a very frequent basis (typically once a week). Moreover, it provides limited clinical improvement and there is a risk of skin cancer due to the use of UV radiation. A psoriasis treatment using CO₂ laser light has also been tried with limited success.

A more recent prior art treatment of psoriasis uses a pulsed dye laser operating at a wavelength of 585nm, with a pulse duration of 0.4msec and fluences in the range of 6 to 10J/cm². While this treatment was generally effective, one significant drawback is the small spot size (of the order of 5mm) of a pulsed dye laser. The small spot size makes the treatment very inefficient since psoriasis typically appears on large areas of the skin. Thus, the treatment becomes a time consuming procedure for the patient. The pulsed dye laser has other shortcomings when used to treat psoriasis related to the fixed and relatively short pulse of this laser, and to the limited degree of tunability of this laser.

Accordingly, a method and apparatus for treating psoriasis that provides effective relief for a relatively long period of time is desirable. Additionally, the method and apparatus will preferably be efficient for treating relatively large areas of skin. Preferably, the treatment will utilize pulsed electromagnetic radiation in the visible and/or near infrared portions of the spectrum and overcome the drawbacks of the prior art treatments.

According to a first aspect of the invention a method for treating psoriasis includes generating one or more pulses of incoherent electromagnetic energy and directing the energy to a region of tissue to be treated. The pulses may have energy fluences in the range of 5 J/cm² to 200 J/cm². In an alternative embodiment the

pulse duration and the number of pulses are selected to control a treatment parameter and optimize the treatment. The parameters that are controlled include the heating of healthy tissue and the penetration depth of the energy. In another alternative the radiation is filtered, thus controlling the radiation spectrum and penetration depth. The filtering may include attenuating an UV portion of the radiation spectrum. In one embodiment a large spot size is created to treat large area psoriasis plaque. A cooling gel is applied to the skin to be treated in another embodiment. The gel may be cooled either before or after it is applied to the skin.

An apparatus for the treatment of psoriatic plaque includes a housing that has a window, in accordance with a second aspect of the invention. A light source that produces incoherent radiation in the visible and near infrared range of wavelength is placed in the housing. A reflector reflects and directs the light through the window to the plaque that must be treated. A filter removes unwanted portions of the spectrum, such as light in the UV range. The filter system may include a fixed filter that attenuates the UV light and a variable filter that attenuates light below a selectable wavelength. Alternatively, the variable filter may pass a selected bandwidth. In another embodiment a light guide is placed between the window and the skin to be treated, and directs the light to the skin. The light guide may be a flexible light guide for a small spot size or a quartz light guide for large spot sizes. Additionally, the light guide may filter a portion of the spectrum. A power supply that includes a pulse generator is provided to power the light source in one embodiment. In another embodiment a microprocessor controls the power supply and the pulse duration and pulse delay. The microprocessor may also include a display for displaying suggested treatment parameters.

A third aspect of the invention is a method of protecting a region of epidermis during a treatment of psoriasis. The protection is effected by applying a gel to the region of epidermis to cool the epidermis. The gel may be cooled before or after it is applied to the epidermis.

Other principal features and advantages of the invention will become apparent to those skilled in the art upon review of the following drawings, the detailed description and the appended claims.

The present invention will now be described further hereinafter, by way of example only, with reference to the accompanying drawings; in which:-

Figure 1 is a schematic illustration of one preferred embodiment of the present invention; and
Figure 2 is a graph of the effective penetration depth of light into bloodless skin (dermis) as a function of wavelength in the range of 400nm to 1000nm.

Before explaining at least one embodiment of the invention in detail it is to be understood that the invention is not limited in its application to the details of construction and the arrangement of the components set forth in

the following description or illustrated in the drawings. The invention is capable of other embodiments or being practiced or carried out in various ways. Also, it is to be understood that the phraseology and terminology employed herein is for the purpose of description and should not be regarded as limiting.

The invention relates to a new method and apparatus for treating psoriasis. Generally, the invention uses pulsed light, preferably incoherent, in the visible range, that causes the clearance of psoriatic plaque. In accordance with one preferred embodiment, the apparatus includes a light source that provides electromagnetic radiation that penetrates into the skin, and reaches depths of the order of 1mm or more. Preferably, the fluence of the light will be sufficient to enable coagulation of the vessels feeding the lesion and the abnormal cells. The light is applied to the skin in pulses to help limit any damage to healthy tissue that surrounds the unhealthy tissue, that might otherwise be caused by heat conductivity. Also, the spot size of the light will preferably be large enough, in the centimeter range, e.g., to enable efficient treatment of large areas of skin.

An apparatus in accordance with one preferred embodiment of the invention includes a high energy, pulsed, incoherent light source, such as a flashlamp. A suitable flashlamp (typically in a linear configuration) will generate pulsed light in the visible and near infrared range of wavelength. The apparatus includes a fixed filter system which cuts off the radiation spectrum, thus reducing the amount of electromagnetic radiation of damaging wavelengths that reaches the skin. Also, a variable filter system that has lower cut off filters that allow radiation above a given wavelength to be transmitted to the skin is provided in one embodiment. Alternatively, the lower cut off filters allow a selected bandwidth of wavelength to be transmitted to the skin. The selection of the variable filter (and thus the wavelength of the transmitted light) will allow the optimal treatment to be provided.

The light is directed to the skin through an opening in a housing that contains the flashlamp and the filter system in one embodiment. In another embodiment a light guide connected to the housing and in contact with, or in the vicinity of, the skin is used to direct the light to the treatment area. Preferably, the system will produce fluences on the skin in the range of up to a few to hundreds of joules per square centimeters in either embodiment.

A coupling gel that can be used in some cases to enhance light coupling to the skin and enable better cooling of the epidermis during treatment.

An apparatus made in accordance with the present invention is useful for treatment of psoriasis. High energy pulsed light can be an effective way of selective photothermolysis of blood vessels and other parts of unhealthy tissue without damage of normal skin. Moreover, an apparatus made in accordance with the present invention is safe and there is little risk of accidental injury

to the operator and patient.

Referring now to Figure 1, a treatment device 10 for treating a region of skin 6 made in accordance with the present invention is schematically shown and includes a flashlamp 1 and a reflector 2 disposed in a housing 5. In the preferred embodiment flashlamp 1 has a linear shape, although other shapes could be used.

Flashlamp 1 may be operated in pulse mode and produce light radiation able to penetrate into a tissue at the millimeter depth or deeper. The pulse rate and delay rate may be selected to provide high radiation density on the treated surface and avoid the overheating of surrounding health tissue due to heat conductivity process. A fluence in the range of from a few joules to more than tens of joules on the surface of skin 6 over an area of several square centimeters, and preferably in the range of from 5 J/cm² to hundreds of J/cm² (200 J/cm², e.g.) is provided by flashlamp 1. The light source will preferably provide a spot size variable from at least some centimeters to some millimeters.

Reflector 2 forms the light beam and reflects it to a light guide 4. In one embodiment reflector 2 may be a metallic reflector. Typically, polished aluminum, which is an easily machinable reflector and has a very high reflectivity in the visible range of the spectrum, can be used. Other bare or coated metals can also be used. Reflector 2 preferably has a shape and cross section to focus light produced by flashlamp 1 at a desired location, such as at the proximal end of light guide 4, or on the surface of skin 6 which is to be treated. One such reflector is described in U.S. Patent No. 5,405,368, which is hereby incorporated by reference.

A filter system 3 is disposed within housing 5, adjacent an opening or window 11. Filter system 3 is thus located in the path of light directed from the flashlamp 1 to the light guide 4 and/or skin 6, and will affect the spectrum of light provided by device 10. Filter system 3 includes one or more fixed filters that attenuate electromagnetic radiation having wavelengths (such as light in the UV range) that will damage the skin and/or overheat shallow layers of skin. Also, in one embodiment, filter system 3 includes one or more variable filters that have lower cut off filters and allow radiation above a given wavelength to be transmitted to the skin. In an alternative embodiment the variable filters transmit a selected bandwidth of wavelength to the skin, thus reducing the need for the fixed filters.

The depth the treatment light penetrates into the tissue or skin 6 is dependent upon the wavelength of the incident light (as will be discussed in greater detail below). Thus, the penetration depth may be controlled by the selection of the filters. Accordingly, the selection of the variable filter (and thus the wavelength of the transmitted light) may be done to optimize the treatment.

Additionally, in an alternative embodiment spectral control is achieved by controlling the parameters of the pulse provided to flashlamp 1. For example, a spectrum with longer wavelengths may be obtained by decreasing

the arc current in the flashlamp.

A power source 8 provides power to flashlamp 1 to produce the pulsed light output. Power source 8 preferably provides single or multiple pulses with delay between pulses which varies from several milliseconds to hundreds of milliseconds. The total fluence to the treated area is the product of the number of pulses and the fluence per pulse and preferably may reach a value of hundreds of joules per square centimeter. The pulse duration can be varied in the range of hundreds of microseconds to tens of milliseconds in the preferred embodiment, and the fluence per pulse is variable in the range of a few joules to hundreds of joules per square centimeters. Power source 8 may include a pulse forming network, such as that shown in U.S. Patent No. 5,405,368, or other circuitry to produce the desired pulses.

Pulse parameters are controlled by a microprocessor based controller 9 in the preferred embodiment. Microprocessor based controller 9 provides the timing functions and prompts the trigger signals that cause power supply 8 to deliver the pulses of power to flashlamp 1. In one embodiment power supply 8 and controller 9 are disposed within housing 5.

In an alternative embodiment microprocessor based controller 9 includes a user interface, such as a display screen and keyboard, buttons, mouse, or other input device, and may be a personal computer. Controller 9 may have information stored therein that aids in the selection of treatment parameters. The physician inputs patient information, such as the depth needed to be treated, and the microprocessor provides suggested treatment parameters, such as wavelength, filter selection, pulse width, and pulse delay. The physician can alter these suggested parameters, but need not refer back to operating guidelines for suggested parameters. This alternative may be used with light sources other than a flashlamp, such as UV or a pulsed laser.

Light guide 4 directs the light to the treatment area on skin 6. If the area to be treated is difficult to access objects or small plaques are to be treated, light guide 4 may be a flexible light guide with the spot size of several millimeters. Alternatively, if large areas are to be treated light guide 4 may be a broad quartz light guide with a spot size of several centimeters, or the use of a light guide may be omitted altogether.

Light guide 4 may be used for spectral control in one alternative embodiment. Spectral control can be achieved by making the light guide from a material that had an absorbing dye dissolved therein. Thus, light transmitted by the light guide will have a spectrum as determined by the absorbing dye.

The treatment is accomplished by coagulation of hemoglobin and overheating of tissue with abnormal pigmentation. However, to avoid overheating of normal epidermis and decrease pain, a transparent gel 7 may be applied on treated skin surface 6.

The cooling time t of an epidermis that has typical dimension d and diffusivity a can be written as:

$$t=d^2/a$$

The epidermis has typical cross dimensions of less than 0.1mm. The diffusivity is approximately $a=3 \times 10^{-7} \text{m}^2 \text{sec}^{-1}$. When a gel is applied the typical cooling time of the epidermis will be on the order of 30msec. Application allows the epidermis to cool during a pulse and to avoid adverse effects if the light pulse duration is approximately the same as the cooling time of the epidermis. In order to increase the cooling effect the gel may be previously cooled down. Alternatively, the gel may be cooled after it is applied to the skin.

As stated above, depth of penetration can be controlled by selection of appropriate wavelength range. The effective depth of penetration into the skin can be estimated by using the effective attenuation coefficient of the dermis that takes into account scattering and absorption of light. As described in S. L. Jacques, Role of Skin Optics in Diagnostic and Therapeutic Uses of Lasers, Springer-Verlag, 1991, pp. 1-21, effective attenuation coefficient of the skin can be written as:

$$\mu_{\text{eff}}=[3\mu_a(\mu_a+\mu_s(1-g))]^{1/2}$$

where μ_a is an absorption coefficient of dermis, μ_s is a scattering coefficient of dermis, g is the anisotropy factor which is defined as the average cosine of the scattering angle for one scattering event. The effective penetration depth which can be estimated from

$$d=1/\mu_{\text{eff}}$$

and is shown in the Figure 2 as a function of wavelength in the range of 400nm to 1000nm. As shown, radiation with longer wavelengths penetrates deeper into the skin than radiation of shorter wavelengths.

The effective penetration depth d is defined as the depth at which the fluence impinging of the skin reaches $1/e$ of the value on the surface of the skin. As shown in Figure 2, penetration goes up by a factor of almost two when the wavelength is increased from 500nm to 600nm. Penetration depths of 2mm can be achieved at a wavelength of 800nm. Thus, in the preferred embodiment the spectrum of the light that reaches skin 6 has a wavelength selectable over the range of 400nm to 1000nm, and particularly from the range of 500nm to 600nm and as high as 800nm. As stated above the spectrum may be controlled using filters, light guides, or pulse widths. Thus, proper filtering and the use of a gel allows selectivity of treatment to be achieved by selection of a desired spectrum of radiation and by cooling shallower skin layers.

Thus, it should be apparent that there has been provided in accordance with the present invention a method and apparatus for treating psoriasis that fully satisfy the

objectives and advantages set forth above. Although the invention has been described in conjunction with specific embodiments thereof, it is evident that many alternatives, modifications and variations will be apparent to those skilled in the art. Accordingly, it is intended to embrace all such alternatives, modification and variations that fall within the spirit and broad scope of the appended claims.

light guide filters a portion of a radiation spectrum.

10. The use of apparatus according to any one of claims 1 to 9 in a method of treatment of psoriatic plaque.

Claims

1. An apparatus for the treatment of psoriatic plaque comprising: a housing (5), including a window (11); a light source (1) disposed within the housing, wherein the light source is capable of producing incoherent radiation in the visible and near infrared range of wavelength; a reflector (2) disposed within the housing and further disposed to direct the light through the window; and a filtering system disposed in a path of the radiation.
2. An apparatus as claimed in claim 1 further including a power supply (8) having a pulse generator in electrical communication with the light source wherein the power supply includes a control input for receiving a plurality of signals that control the pulse generator, and further including a microprocessor (9) having a control output in electrical communication with the control input.
3. An apparatus as claimed in claim 2 wherein the microprocessor includes means for displaying suggested treatment parameters.
4. An apparatus as claimed in any one of claims 1, 2 or 3 wherein the filtering system includes a fixed filter that attenuates radiation in an UV range.
5. An apparatus as claimed in any one of claims 1 to 4 wherein the filter system includes a variable filter that attenuates radiation below a selectable wavelength.
6. An apparatus as claimed in any one of claims 1 to 5 wherein the filter system includes a variable filter that attenuates the radiation outside of a selected bandwidth.
7. An apparatus as claimed in any one of claims 1 to 6 including a light guide disposed adjacent the window and capable of directing the radiation to a treatment region.
8. An apparatus as claimed in claim 7 wherein the light guide is a quartz light guide.
9. An apparatus as claimed in claim 7 or 8 wherein the

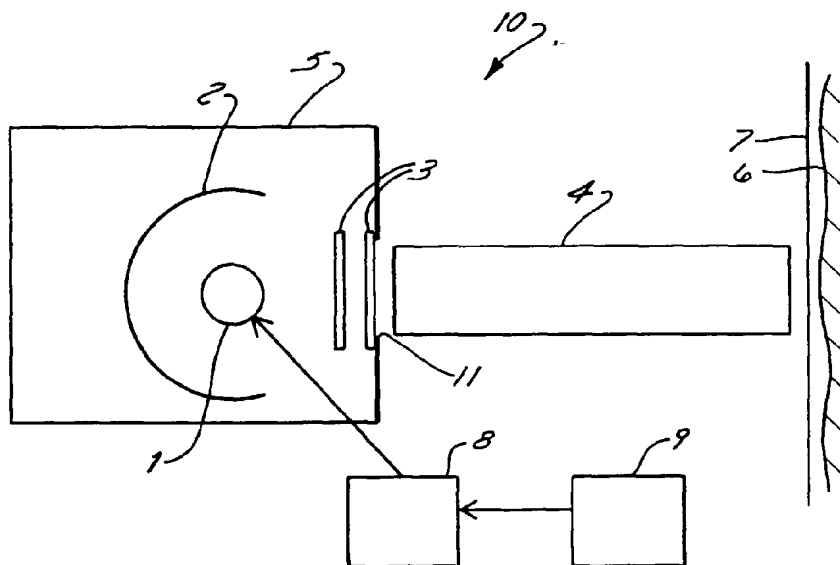


FIG. 1

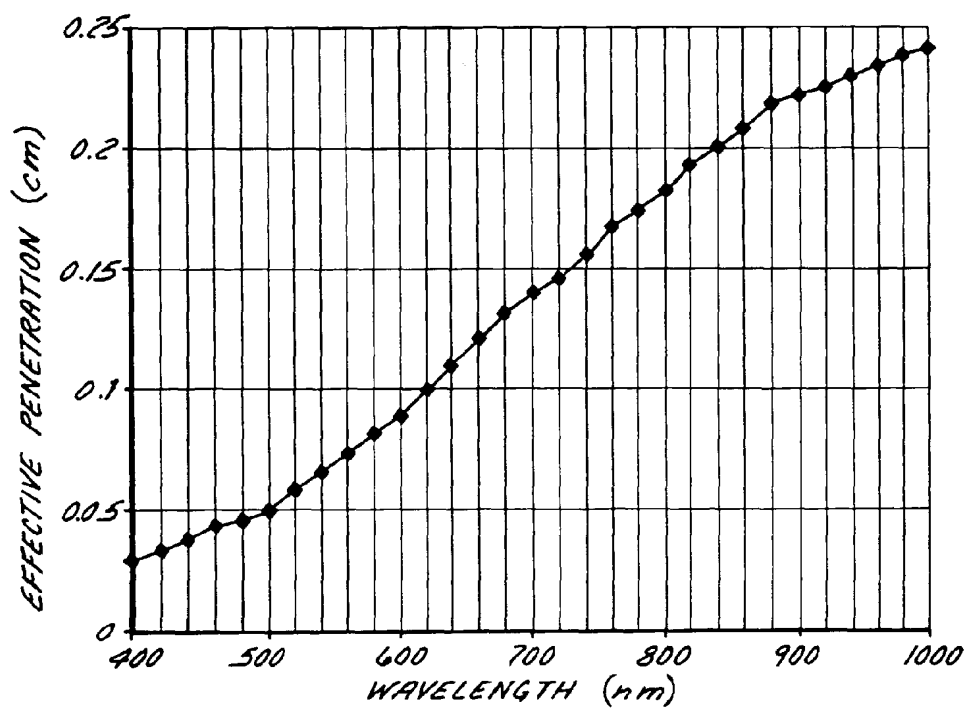


FIG. 2



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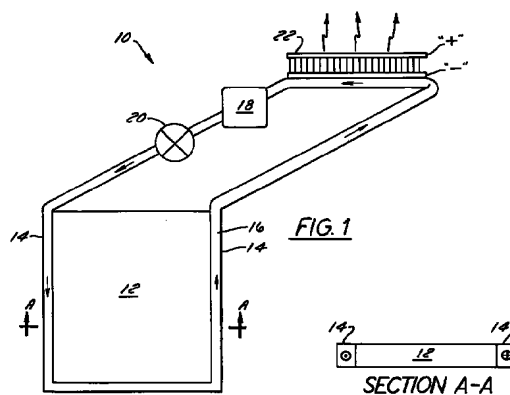
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(54) Method and apparatus for controlling the thermal profile of skin

(57) A method is provided for cooling skin during irradiation treatment including thermally coupling a window (12) to the skin and actively extracting heat from the window. The window may be transparent to therapeutic wavelengths transmitted to and through the window and to the skin below. An apparatus for therapeutic treatment of skin tissue is provided, including a source (24) of electromagnetic radiation, a window (12) transparent to the radiation coupled to the skin being treated to remove heat from the skin, coolant fluid (16) coupled to the window to extract heat from the window, and a heat exchanger (22/40) thermally coupled to the coolant to remove thermal energy from the coolant.



EP 0 783 904 A2

Description

This invention relates to a method and apparatus for controlling the thermal profile of skin. More particularly, it relates to a method of reducing the temperature of an outer layer of skin while the temperature of an inner layer of skin is elevated.

Electromagnetic radiation is used to treat a variety of skin disorders, such as vascular and pigmented lesions, hair removal, skin rejuvenation, psoriasis, among others. This radiation is typically applied to the surface of the skin from a variety of radiation sources, such as lasers that emit coherent light, flashlamps emitting incoherent light and microwave radiation sources, among others. Whatever the source of electromagnetic radiation, in order to provide treatment without damaging the epidermis and surrounding tissue, careful consideration must be given to the problem of maintaining the proper thermal profile in the skin.

For example, a method called selective photothermolysis uses selective absorption of pulses generated in the visible and near-visible ranges of electromagnetic spectrum to produce selective thermal injury to the skin. In this treatment, the skin is exposed to short pulses of electromagnetic radiation to heat tissue chromophores and blood vessels. Short pulses of intense radiation are necessary to transmit energy to the target tissue at a level that can damage the tissue before it can be cooled off. Since the cooling time for the epidermis is typically around 9 milliseconds, the pulses must provide extreme localized heating. Repeated pulses of a few milliseconds duration followed by delays of a few tens of milliseconds provide optimal deep skin heating while minimizing damage to outer skin layers. By controlling the pulse width, pulse delay, energy per pulse, and the frequency of radiation applied to the skin, the temperature distribution of the skin as a function of depth -- the thermal profile of the skin -- can be controlled to prevent damage to the skin, while providing enough thermal energy for treatment.

As the depth of the tissue to be treated increases, so does the need to cool the outer layers of skin to prevent injury. Therefore, when treating skin greater than one millimeter deep, positive cooling of the epidermis may be required. There are three basic methods employed to cool the epidermis: cooling using a layer of radiation transparent gel; cooling using "ice cubes"; and cryogen spurt cooling.

In the first of these methods, cooling with a gel, a pre-cooled transparent gel is applied to the surface of the skin to conduct heat away from the epidermis into the gel. This method is limited by its ability to reduce the epidermis temperature by no more than about 20° Celsius. This limited cooling may not be sufficient if the treatment includes intense heating. In addition, it is cumbersome to apply the gel during treatment while simultaneously irradiating the skin. A further drawback is the passive nature of the cooling: heat is extracted from the skin into a precooled material in contact with

the skin that heats up as the skin cools down. As thermal energy is conducted into the gel from the skin, the gel heats up until it reaches a temperature near body temperature. No method is provided to actively extract heat from the gel itself thereby maintaining it at a low temperature.

The second of these methods, cooling using "ice cubes", involves placing thin transparent ice cubes, approximately 5-7 millimeters thick, in contact with the skin. Applying the ice cubes and maintaining the proper contact with the skin is also cumbersome and difficult. Timing the cooling relative to the radiation pulses is also difficult to control. This method, too, is passive, since the heat transfer is limited to the thermal capacity of the ice itself. No means for actively extracting heat from the ice in contact with the skin is provided.

The third method, cryogen spurt cooling, involves spraying the surface of the skin being treated with a refrigerant, such as R-12, that evaporates at room temperature and pressure. The refrigerant is sprayed on the epidermis in pulses that typically vary between 5 and 80 milliseconds in duration. These pulses cool a surface area of skin of about seven millimeters in width. Since this method involves spraying a pressurized liquid coolant on the skin, the timing of cooling with respect to treatment irradiation is more controllable than the foregoing methods. A further advantage is the amount of cooling possible using this method; commonly used refrigerants can cool the epidermis as much as 40° Celsius. Drawbacks to this method include difficulty in controlling the amount of cooling, the inability to cool more than a small area of skin, and the difficulty in properly aligning the cooling and heating mechanisms.

The foregoing illustrates the need for a new method of cooling the skin that is more convenient, provides better control of temperature and timing, and is capable of cooling a larger surface area of skin.

The present invention is directed to a method of cooling skin as protection against thermal damage during radiation treatment, and includes the steps of thermally coupling a window to the skin, and actively extracting heat from the window. To extract heat, a coolant fluid can be thermally coupled to the window and its heat may be conducted into the coolant fluid. The fluid, in turn, may be conducted away from the window by a pump, for example, and its flow may be regulated by a valve. Thermal energy may be removed from the coolant fluid, such as by coupling the fluid to a heat exchanger, for example, a thermoelectric cooler. Skin temperature can be monitored, such as by transmitting infrared radiation from the skin through the window, then electronically sensing this radiation. A thermally conductive gel may be interposed between the skin and the window, the window pressed against the skin, and a portion of the gel extruded from between the window and the skin. This gel may be a water-based thermally conductive gel, and may contain antifreeze.

The present invention is also directed to a method for controlling the depth-wise temperature distribution of

skin tissue by thermally coupling skin tissue to a window that is transparent to therapeutic wavelengths of electromagnetic radiation, applying therapeutic wavelengths of radiation to the window, transmitting the wavelengths through the window, and applying the wavelengths to the skin tissue. The method may also include extracting thermal energy from the skin tissue by cooling the window a predetermined period before the radiation is applied to and transmitted through the window and applied to the skin.

In addition, the present invention is directed to an apparatus for therapeutic treatment of skin tissue including a source of therapeutic radiation, a window transparent to the radiation, coolant fluid thermally coupled to the window and adapted to remove thermal energy from the window, and a heat exchanger thermally coupled to the coolant to remove thermal energy from the coolant. The window may have a crystalline structure, such as sapphire or quartz. It may be a synthetic sapphire. It preferably transmits radiation in the 0.3 to 4.5 micron band of radiation. The window's thermal conductivity is preferable at least 10 W/m²°C. More preferable it is at least 25 W/m²°C. Most preferable it is at least 40 W/m²°C. A conduit may be thermally coupled to the window and adapted to convey the coolant into thermal contact with the window. The coolant flow may be regulated, such as by a valve. A coolant moving element may be provided to propel the coolant through the conduit, such as a pump. The radiation source may emit incoherent electromagnetic radiation, such as a flashlamp, or may emit coherent radiation, such as a laser. The coolant may be thermally coupled along a lateral edge of the window. The window itself may transmit the coolant fluid, and may transmit fluid through the radiation path. The present invention may include a heat exchanger thermally coupled to the window to remove thermal energy from the window, and cooling fluid coupled to the heat exchanger to remove heat from the heat exchanger. The heat exchanger may be a thermoelectric cooler, and the coolant fluid may be air or a liquid. If the coolant fluid is a liquid, it is preferable maintained in thermal contact with the heat exchanger by a liquid conduit.

The present invention will now be described further hereinafter, by way of example only, with reference to the accompanying drawings; in which:-

FIGURE 1 is an illustration of an apparatus for cooling epidermis;

FIGURE 2 is a cross-sectional view of a device for the treatment of lesions including the apparatus of **FIGURE 1** thermally coupled to the skin at a treatment site;

FIGURE 3 is a cross-sectional view of a window with a thermally coupled heat exchanger and heat sink; and

FIGURE 4 is a cross-sectional view of a window with a thermally coupled heat exchanger and liquid conduit.

Before explaining at least one embodiment of the invention in detail it is to be understood that the invention is not limited in its application to the details of construction and the arrangement of the components set forth in the following description or illustrated in the drawings. The invention is capable of other embodiments or being practiced or carried out in various ways. Also, it is to be understood that the phraseology and terminology employed herein is for the purpose of description and should not be regarded as limiting.

The present invention is directed to a method and apparatus for cooling skin during electromagnetic irradiation treatments. Regarding **FIGURE 1**, a cooling system 10 includes a transparent window 12 thermally coupled to a cooling channel 14. Cooling channel 14 contains a coolant 16, which is propelled through channel 14 by coolant pump 18, and is also regulated in its flow by coolant valve 20. A heat exchanger 22 is provided to regulate the temperature of coolant 16.

The transparent window serves two primary purposes in the present invention: conducting heat away from the skin, and transmitting electromagnetic radiation from a radiation source (not shown) to the skin for treatment. It is "transparent" in the sense that it allows desired therapeutic wavelengths of electromagnetic radiation to substantially pass from the radiation source to the area of skin being treated. To provide superior heat conduction and high radiation transmissivity, window 12 has a thermal conductivity of at least 10 W/m²°C is preferred to provide the necessary cooling capacity for typical applications. A material having a thermal conductivity of at least 25 W/m²°C (such as synthetic sapphire) is more preferred. Several commonly available crystalline materials such as quartz or sapphire provide even better cooling, and have thermal conductivities of around 40-45 W/m²°C. From a technical perspective, the best window material currently known is diamond, which has a thermal conductivity of around 900 W/m²°C. Diamond windows are currently impractical due to cost, however.

Sapphire's applicability is enhanced by its relative transparency to radiation in a band extending from 0.3 to 4.5 μm, enabling its use with a variety of coherent and incoherent radiation sources. The operation of a sapphire window is highly efficient. For example, a sapphire window measuring 35 mm by 8 mm and having a 5 mm thickness can be cooled to 0° Celsius from room temperature in less than 20 seconds, so that its temperature distribution will be highly uniform.

Cooling channel 14 is designed to contain coolant 16 and to thermally couple it to window 12, thus allowing heat transmitted from the skin to the window to be transferred into the coolant for removal from the treatment site. In the preferred embodiment, and as shown in **FIGURE 1**, cooling channel 14 is formed along a lateral edge of window 12. Alternatively, cooling channel 14 may be integrally formed within window 12 itself. If a portion of cooling channel 14 passes through window 12 and coolant 16 is transparent to the therapeutic radi-

ation, therapeutic radiation can be directed through both window 12 and coolant 16 to irradiate the treatment site.

Coolant 16 typically includes a high heat capacity fluid such as water. In this embodiment, it removes heat from the skin by conduction from the window. To transfer thermal energy from the treatment site to the window, the window must be maintained at a temperature below that of the treatment site. To transfer thermal energy from the window to the coolant, the coolant must be maintained at a temperature below that of the window. To significantly limit damage to the skin during treatment, the treated skin should preferably be cooled to a temperature at or near the freezing point of water. To provide a temperature this low, the coolant must remain fluid at a temperature of 0° Celsius or below. Thus the coolant preferably includes an antifreeze. If the coolant is conducted through channels in the window itself, rather than along the lateral edges of the window, it should preferably be transparent to radiation in the band of therapeutic wavelengths.

Coolant pump 18 is provided to pump coolant 16 through cooling channel 14 in the direction of the arrows shown in FIGURE 1, and valve 20 regulates the flow of coolant through the cooling channel. Cooling can be controlled by regulating the output of the pump, typically by regulating pump speed or capacity, regulating the amount of flow restriction provided by the cooling valve or regulating both the pump and the cooling valve simultaneously. Preferred pumps include reciprocating, centrifugal and peristaltic pumps.

In the preferred embodiment, heat exchanger 22 is provided to transfer heat energy to a secondary cooling fluid such as air or water. The heat exchanger illustrated here is a thermoelectric cooler which, due to its relatively small size and low power consumption, is particularly suited to removing heat energy from the coolant. In this embodiment, the coolant is in a closed loop, picking up heat at the window and releasing heat in the heat exchanger. By regulating the current flowing through the thermoelectric cooler (which is produced in the embodied heat exchanger by application of a voltage to its terminals labeled "+" and "-") the amount of heat extracted from the coolant can be varied. In an alternative embodiment, rather than providing a closed loop for the primary coolant, a large reservoir of chilled coolant can be provided to cool the skin. In this alternative embodiment, the heat exchanger would be replaced with a large precooled reservoir or tank of coolant. Fluid would be pumped out of this reservoir and through the window.

FIGURE 2 discloses another embodiment of the invention including a radiation source 24 for emitting electromagnetic radiation, a reflector 26 for reflecting the radiation toward a treatment site 28, a light guide 30 for directing and transmitting the radiation toward treatment site 28, a radiation filter 32 for restricting the radiation transmitted to therapeutic bands of frequencies, window 12 for transmitting the therapeutic wavelengths toward treatment site 28 while simultaneously conducting thermal energy from the skin, a gel 34 for transmit-

ting the therapeutic radiation wavelengths while simultaneously conducting thermal energy from the skin to window 12, a radiation sensor 36 for sensing the temperature of treatment site 28, cooling channels 14 and pressure sensor 38. The path of therapeutic radiation emitted from radiation source 24 to the treatment site is shown schematically as dashed lines. Coolant passing through the cooling channels can be conveyed, controlled and cooled in a manner similar to that disclosed in FIGURE 1 and the accompanying text.

Radiation source 24 provides pulsed electromagnetic radiation including therapeutic wavelengths of radiation. In this embodiment, radiation source 24 is a flash lamp that emits incoherent radiation in a broad spectrum. Alternatively, a radiation source capable of providing coherent radiation, such as a laser radiation source, may also be effectively employed.

Reflector 26 is preferably polished metal, for example polished aluminum, to reflect at least the radiation in therapeutic wavelengths.

Light guide 32 is employed to gather and direct radiation from the radiation source to treatment site 28. Typically an optical fiber is employed for radiating relatively small treatment sites, and a quartz light guide is employed for radiating larger treatment sites.

Radiation filter 32 is employed to filter out unwanted wavelengths of radiation, typically wavelengths that are harmful to the skin, such as radiation in the ultraviolet spectrum. One or more filters may be employed to transmit a band of wavelengths that are tailored to penetrate the skin to a predetermined depth.

A gel 34 may be disposed between the surface of the skin and the window, thermally coupled to both, to provide better transmission of therapeutic wavelengths by reducing backscatter off the surface of the window in proximity to the skin and to provide more effective conduction of thermal energy from the skin to the window by eliminating pockets of air that may remain between skin at the treatment site and window 12. Skin at the treatment site is often rough and uneven. Consequently, pockets of air (not shown) may remain between window 12 and skin at the treatment site preventing good thermal contact when window 12 is pressed against the skin. Heating and cooling may be uneven and unpredictable. A wetting agent, such as gel 34, when applied between window 12 and skin at the treatment site, fills surface imperfections on the skin, and allows air to be expelled when window 12 is pressed against the skin. Water-containing gels are particularly effective due to their relatively high thermal conductivity, their ability to wet both the treatment site and the window, and their ability to transmit a wide range of therapeutic radiation wavelengths. In use, a gel is applied to the window or the skin in quantities greater than that needed to fill the surface imperfections, then the window is brought into contact with the skin. As pressure is applied to the window, excess gel (along with entrained air) is squeezed out along the sides of the window, providing a thin, thermally conductive layer of gel between the window and

the treatment site.

Radiation sensor 36 may be employed to produce a signal indicative of the degree of skin heating or cooling. A sensor responsive to infrared wavelengths of light emitted by skin at the treatment site is particularly well suited to this application. The window is preferably oriented between the sensor and the treatment area to pass thermal radiation emitted from the skin to the sensor. Radiation sensors that are responsive to radiation in the 2 to 5 micron band are particularly suitable for measuring radiation emitted from the skin.

Cooling channels 14 may be employed as described above in accordance with the description in accompanying FIGURE 1.

In order to achieve good thermal contact with the skin surface, it is preferable to apply and maintain pressure on the window against the treatment site. Pressure sensor 38 is disposed to sense this pressure. In this embodiment, it is fixed with respect to the window. Since the pressure sensor indicates contact between the window and the treatment site, it can be monitored to indicate the onset of skin cooling and thus to control a delay (if desired) between cooling and generation of the therapeutic radiation. For example, epidermis at a 50 micron depth can be cooled from 32° Celsius to 6° Celsius in 1 second. By delaying the light pulse for 1 second after cooling initiation, significant damage to the epidermis layer can be avoided. Window temperature is also a factor in determining the light pulse delay. For example, if the window is cooled to -15° Celsius before it is applied to the skin, the time to cool the epidermis at a depth of 100 microns is reduced to 0.1 seconds. By monitoring the pressure sensor, and controlling the temperature of the window, optimal cooling can be provided.

FIGURE 3 discloses a cross-sectional view of window 12, thermally coupled to heat exchanger 40, with attached heat sink 42. A fan 44 is provided to blow air across heat sink 42. In this embodiment, unlike the embodiment shown in FIGURES 1 and 2, heat exchanger 40 (here shown as a thermoelectric cooler) is thermally coupled to window 12 directly, eliminating the intermediate coolant shown in FIGURE 1 and indicated in FIGURE 2. This apparatus is preferred when a quick response is required and it is possible to apply a large window to the treatment site. By thermally coupling heat exchanger 40 directly to window 12, it can cool the window directly. In comparison, the apparatus disclosed in FIGURE 1 allows heat exchanger 22 to cool the window by first chilling coolant 16, which in turn cools window 12. In such an embodiment, both the mass of coolant 16 and the mass of window 12 must be chilled to a temperature below the desired skin temperature to effect cooling to that temperature. By placing heat exchanger 40 in direct thermal contact with the window 12, only the mass of the window need be cooled to a temperature below the desired skin temperature to effect cooling to that temperature. This direct coupling allows more rapid window cooling and more precise

control of skin temperature. This improved cooling and control, in turn, provides for more rapid cycling of the radiation source (shown in FIGURE 2) and shorter treatments. Heat exchanger 40 is preferably controlled similarly to the heat exchanger of FIGURE 1.

To provide more rapid cooling and to increase the efficiency of heat exchanger 40, it may be thermally coupled to heat sink 42. A preferred material for the heat sink is aluminum or another high conductivity metal. Fins may be provided on heat sink 42 to enhance cooling.

Fan 44 may also be employed to increase the air flow about heat exchanger 40 to transfer heat away from the hot side of heat exchanger 40 or the fins of heat sink 42.

Since typical heat exchangers are not transparent to the common wavelengths of therapeutic radiation used for treatment, the heat exchanger or exchangers are preferably thermally coupled to the window along its lateral edges, thereby providing a path for the therapeutic radiation to reach the treatment site. Alternatively, a window larger than the treatment site can be provided, and the heat exchanger or exchangers can be coupled to an upper surface of the window adjacent to, but not obstructing, the path of the therapeutic radiation that passes through the window and impinges upon the skin.

FIGURE 4 discloses an alternative embodiment of window 12 and thermally coupled heat exchanger 40 similar to that of FIGURE 3. In the FIGURE 4 embodiment, fluid coolant flow (such as coolant 16, above) through cooling channel 46 removes heat from heat exchanger 40. Cooling channel 46 is thermally coupled to heat exchanger 40 to remove thermal energy from heat exchanger 40. The coolant may be drawn from an external reservoir of fluid. Heat exchanger 40 in this embodiment is a thermoelectric cooler capable of pumping heat from a cool side to a hot side of the device. Since heat is pumped from window 12 to cooling channel 46, the coolant in cooling channel 46 need not be maintained at a temperature below the target temperature of the skin, as is the case in the FIGURE 1 embodiment. Thus, sources of coolant fluid warmer than that used in the FIGURE 1 embodiment, such as a cold water tap, may be sufficient for many applications.

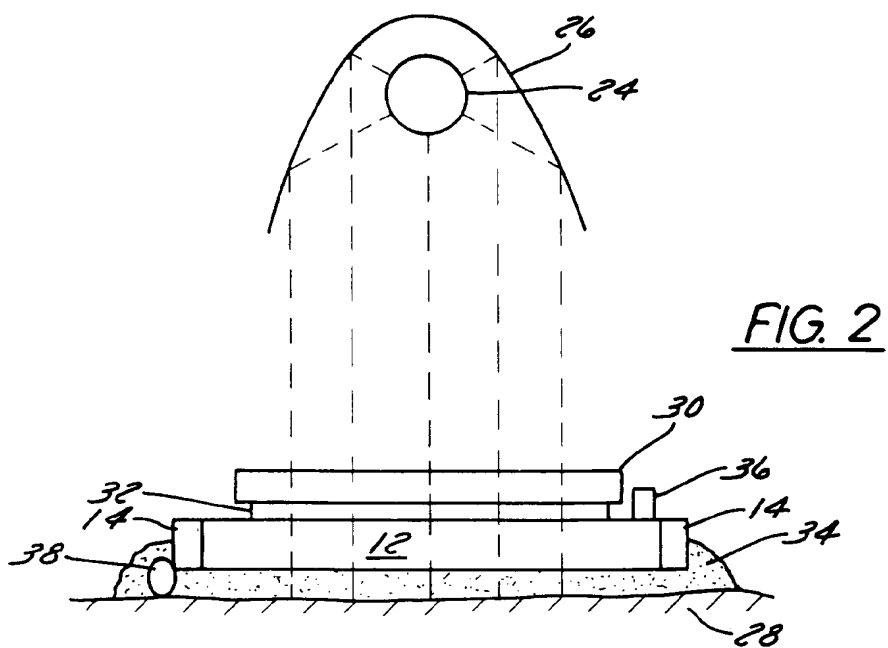
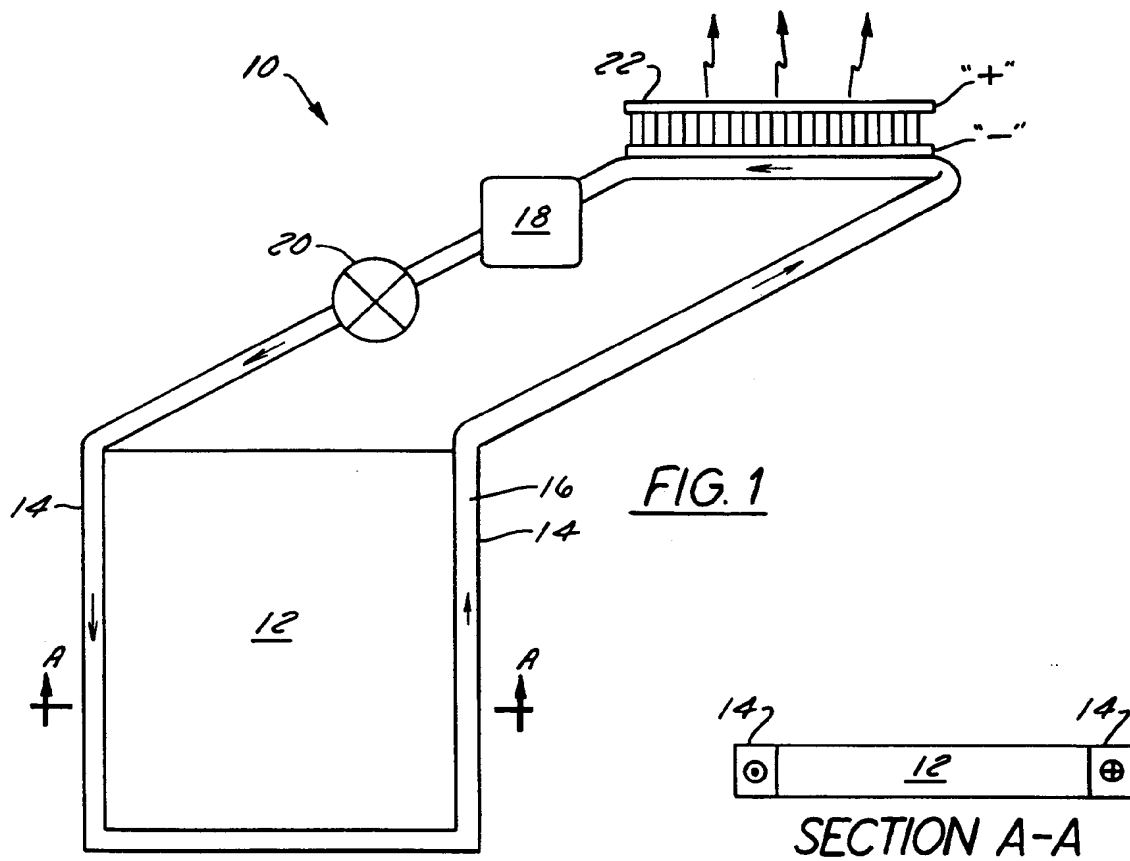
In an alternative embodiment, the window 12, heat exchanger 40 and cooling channel 46 of the FIGURE 4 device can replace the window 12 and cooling channel 14 of the FIGURE 1 device, thereby providing a system with two heat exchangers and an intermediate coolant. This will provide superior temperature control and a faster cooling response time than that provided by the FIGURE 1 or FIGURE 4 embodiments alone.

Thus, it should be apparent that there has been provided in accordance with the present invention a method and apparatus for cooling skin and the selective heating of lesions that fully satisfies the objectives and advantages set forth above. Although the invention has been described in conjunction with specific embodiments thereof, it is evident that many alternatives, mod-

ifications and variations will be apparent to those skilled in the art. Accordingly, it is intended to embrace all such alternatives, modifications and variations that fall within the spirit and broad scope of the appended claims.

Claims

1. A method of cooling skin to protect said skin from thermal damage during irradiation treatment using therapeutic bands of electromagnetic radiation characterised by the steps of:
 - thermally coupling a window (12) to said skin, actively extracting heat from said window.
2. A method as claimed in claim 1, wherein the step of actively extracting heat from said window includes the steps of:
 - thermally coupling a coolant fluid (16) to said window;
 - conducting heat from said window to said coolant fluid; and
 - conducting said coolant fluid away from said window after said step of conducting heat from said window to said coolant fluid.
3. A method as claimed in claim 2, further comprising the step of:
 - removing thermal energy from said coolant fluid.
4. A method as claimed in claim 3, wherein the step of removing thermal energy from said coolant fluid further includes the step of thermally coupling said coolant fluid to a heat exchanger.
5. A method as claimed in any one of claims 1 to 4, further comprising the step of monitoring the temperature of said skin.
6. A method as claimed in claim 5, wherein the step of monitoring further comprises the steps of:
 - transmitting radiation emitted by the skin through said window;
 - electronically sensing said emitted radiation, wherein the step of transmitting radiation occurs before said step of electronically sensing.
7. A method as claimed in any one of claims 1 to 6 wherein the step of thermally coupling a window to the skin includes the step of:
 - interposing a thermally conductive gel (34) between said window and said skin.
8. An apparatus for therapeutic treatment of skin tissue comprising:
 - a source (24) of therapeutic electromagnetic radiation having a path of radiation;
 - a window (12) transparent to said therapeutic radiation and intersecting said path of radiation; and
 - a means (14,16,22,40) for actively extracting thermal energy from said window.
9. An apparatus as claimed in claim 8, wherein said means for actively extracting heat from said window comprises:
 - coolant fluid (16) thermally coupled to said window and adapted to remove thermal energy from said window;
 - a heat exchanger (22) thermally coupled to said coolant fluid and adapted to remove thermal energy from said coolant fluid; and
 - a conduit (14) thermally coupled to said window and adapted to convey said coolant fluid into thermal contact with said window.
10. An apparatus as claimed in claim 9 further comprising a coolant flow regulating means (20) in fluid communication with said conduit to regulate the flow of said coolant fluid.
11. An apparatus as claimed in any one of claims 9 or 10, wherein the coolant fluid is thermally coupled to the window along a lateral edge of the window.
12. An apparatus as claimed in any one of claims 9 or 10, wherein the window is adapted to transmit the coolant fluid.
13. An apparatus as claimed in claim 12, wherein the window is adapted to transmit coolant fluid through the radiation path.
14. An apparatus as claimed in claim 8 or any one of claims 10 to 13 when appendent to claim 8, wherein said means for actively extracting heat from said window includes:
 - a heat exchanger (22) thermally coupled to said window and adapted to remove thermal energy from said window; and
 - coolant fluid (16) thermally coupled to said heat exchanger and adapted to remove thermal energy from said heat exchanger.
15. An apparatus as claimed in any one of claims 8 to 14, wherein said window has a thermal conductivity of at least 10 W/m²°C.



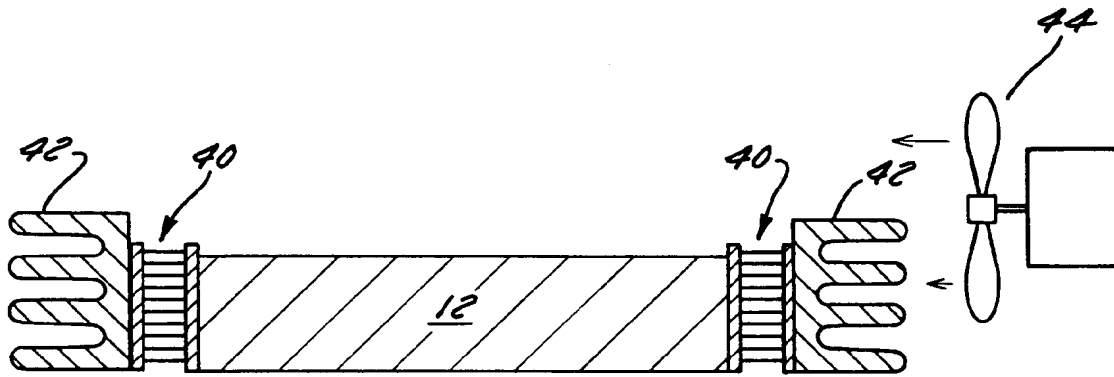


FIG. 3

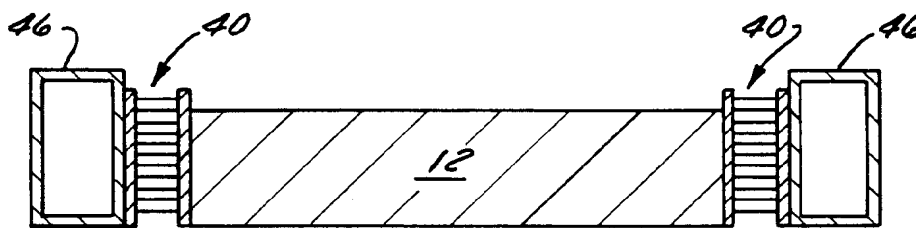





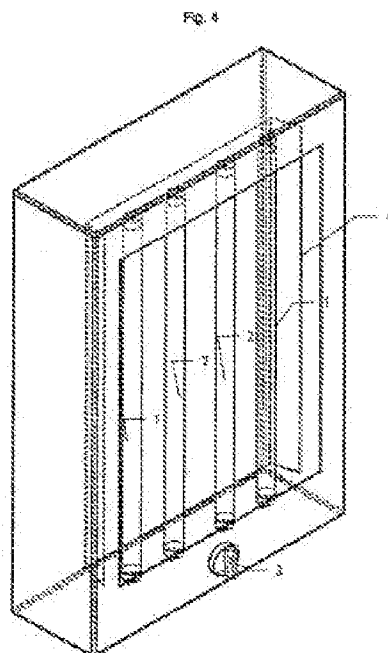


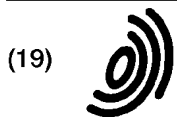
FIG. 4

Device of cosmetical treatment of vulgaris acne**Publication number:** EP0884066 (A2)**Publication date:** 1998-12-16**Inventor(s):** KOEHLER WOLFGANG [DE]**Applicant(s):** SLI LICHTSYSTEME GMBH [DE]**Classification:****- international:** **A61N5/06; A61N5/06;** (IPC1-7): A61N5/06**- European:** A61N5/06C2; A61N5/06W6**Application number:** EP19980110195 19980604**Priority number(s):** DE19971024299 19970609**Also published as:** EP0884066 (A3) EP0884066 (B1) DE19724299 (A1)**Cited documents:** US5549660 (A) DE4143168 (A1)**Abstract of EP 0884066 (A2)**

the method involves irradiating the affected parts with light with two emission spectra, one in the blue region from 400 nm to 45 nm and the other in the red region from 580 to 630 nm. The irradiation is performed once per day for 15 minutes. The method is implemented using a light with a housing contg. at least one lamp and at least one reflector behind the lamp. The housing also contains a switching timer for setting and limiting the exposure time.



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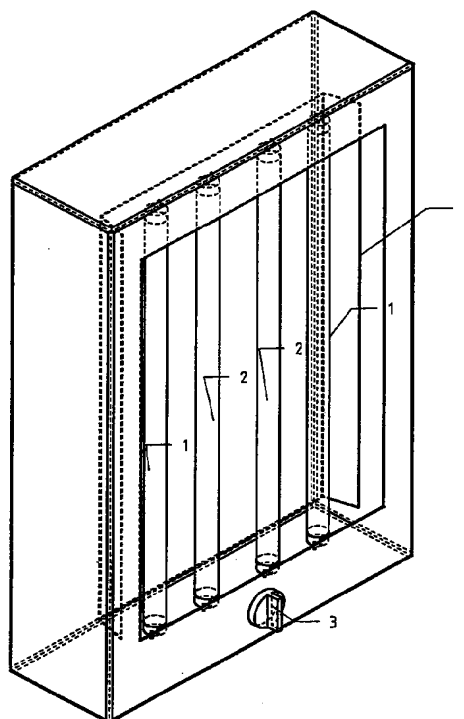
Bemerkungen:

Ein Antrag gemäss Regel 88 EPÜ auf Berichtigung der Zeichnungen 4 und 5 liegt vor. Über diesen Antrag wird im Laufe des Verfahrens vor der Prüfungsabteilung eine Entscheidung getroffen werden (Richtlinien für die Prüfung im EPA, A-V, 3.).

(54) Verfahren und Vorrichtung zur kosmetischen Behandlung von Akne Vulgaris

(57) Bei einem Verfahren zur kosmetischen Behandlung von Akne vulgaris durch Bestrahlung der betreffenden Partien wird Licht mit zwei Emissionsspektren verwendet, von denen das eine im blauen Bereich (A) von 400 nm bis 450 nm und das andere im roten Bereich (B) von 580 nm bis 630 nm liegt. Die Bestrahlung erfolgt mit einer Leuchte (1, 2) in einem Gehäuse (3), in dem zumindest eine Lampe (5, 6, 7, 8; 9, 10) und zumindest ein Reflektor (16) dahinter angeordnet sind, sowie eine Schaltuhr (4) zur Einstellung und Begrenzung der Bestrahlungszeit.

Fig. 4



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Beschreibung

Die Erfindung betrifft ein Verfahren zur kosmetischen Behandlung von Akne Vulgaris durch Bestrahlung der betreffenden Partien mit Licht, sowie eine Vorrichtung zur Ausführung dieses Verfahrens.

Es ist ein Verfahren dieser Art bekannt, bei welchem zur Bestrahlung speziell des Gesichts UV-Licht verwendet wird. Dabei ist jedoch nachteiligerweise eine Erythembildung möglich, ferner eine unerwünschte Oxidation der Hautpigmente.

Ferner ist eine Behandlung mit Creme bekannt, die etwa 0,5 % Benzoylperoxid enthält. Der Nachteil dieser Behandlung besteht in Hautaustrocknung.

Die der Erfindung zugrundeliegende Aufgabe wird darin gesehen, ein Verfahren der eingangs genannten Art nebst einer Vorrichtung zur Ausführung desselben zu schaffen, wodurch nicht nur die genannten Nachteile vermieden werden, sondern darüberhinaus sich ein besonders guter kosmetischer Effekt erzielen läßt.

Diese Aufgabe wird nach der Erfindung verfahrensmäßig dadurch gelöst, daß Licht mit zwei Emissionsspektren verwendet wird, von denen das eine im blauen Bereich von 400 nm bis 450 nm und das andere im roten Bereich von 580 nm bis 630 nm liegt.

Die erfindungsgemäße Bestrahlung ergibt weder UV-Schäden noch signifikante Hautaustrocknung.

Erfindungsgemäß addieren sich die beiden Emissionsspektren zu einem Gesamtspektrum, das dem Aktionsspektrum der Inaktivierung des Propionibacterium acnes angepaßt ist und mit einem biostimulierenden Effekt auf die Zellen der Haut wirkt. Das Propionibacterium acne enthält nämlich Porphyrine, die durch Licht großer Wellenlänge erregt werden können, was zum Tod des Bacteriums führt.

Es wurde eine erfindungsgemäße Bestrahlung mit die genannten Emissionsspektren aufweisendem, blaurotem Mischlicht mit einer Bestrahlung mit blauem Licht und mit einer Bestrahlung mit weißem Licht verglichen, ferner auch mit einer Behandlung mit der genannten Benzoylperoxid-Creme, und zwar an 61 Probanden, die leichte bis mittlere Akne aufwiesen.

Die Probanden waren instruiert worden, jeden Tag eine Bestrahlung von 15 Minuten Länge mit entsprechenden Leuchten durchzuführen, bzw. die genannte Creme zweimal täglich aufzutragen.

Die Auswertung an den Probanden erfolgte alle vier Wochen. Die folgende Tabelle zeigt das Ergebnis dieser Auswertung:

Beobachtung	Blau / rotes Licht		Blaues Licht		Weißes Licht		Creme	
	Arzt %	Patient %	Arzt %	Patient %	Arzt %	Patient %	Arzt %	Patient %
Schlechter / Unverändert	27	27	25	50	46	46	19	19
Leichte / bescheidene Besserung	18	27	42	33	46	46	44	50
markante Besserung	55	46	33	17	8	8	37	31

Die unter "Arzt" aufgeführte Zahl entspricht jeweils der Beurteilung durch einen Arzt, die unter "Patient" der des Probanden nach der Bestrahlungs- bzw. Cremekur.

Wie man der Tabelle entnimmt, wurden die besten Resultate mit dem erfindungsgemäßen blauroten Mischlicht erzielt, und zwar mit einer mittleren Reduktionsrate von 66 % bei entzündlichen und 42 % bei nichtentzündlichen Läsionen, während blaues Licht diesbezüglich 50 % und 32 %, weißes Licht 21 % und 0 %, und Benzoylperoxid-Creme 61 % und 58 % ergab.

Die ärztliche Auswertung ergab eine markante Besserung bei 55 % der Fälle unter Verwendung des erfindungsgemäßen rotblauen Mischlichts, bei 33 % der Fälle bei blauem Licht, bei 8 % der Fälle bei weißem Licht und bei 37 % der Fälle bei Verwendung der Benzoylperoxid-Creme.

Die Auswertung durch die Probanden selber zeigte jeweils markante Verbesserungen mit 46 % bei Verwendung des erfindungsgemäßen rotblauen Mischlichts, von 16 % bei blauem Licht, von 8 % bei weißem Licht und von 31 % bei Cremebehandlung. Trockenheit der Haut wurde speziell bei der Cremebehandlung festgestellt. Bei der Lichtbehandlung war sie demgegenüber geringfügig.

Figur 1 zeigt die spektrale Energieverteilung einer Lampe mit blauem Leuchtstoff, während Figur 2 dieselbe bei einer Leuchtstofflampe mit rotem Leuchtstoff darstellt. Wie man feststellt, befinden sich die erfindungsgemäßen Emis-

sionsspektren jeweils innerhalb des Bereichs der hauptsächlichsten spektralen Energie.

Figur 3 zeigt das Aktionsspektrum der Inaktivierung des *Propionibacterium acnes*.

Im folgenden werden anhand der Fig. 4 und 5 bevorzugte Ausführungsformen von Lampen beschrieben, deren Emissionsspektren den Diagrammen nach den Fig. 1 und 2 entsprechen, so daß sich mit denselben die beschriebenen kosmetischen Behandlungen mit blaurotem Licht nach der Erfindung durchführen lassen. Mit einer Leuchte entsprechend der Fig. 4 wurden die Versuche durchgeführt und die in obiger Tabelle unter "Blau/rotes Licht" angegebenen Werte ermittelt.

Beide erfindungsgemäßen Leuchten 1 bzw. 2 besitzen ein hier rechteckiges Gehäuse 3, in dem zumindest eine Lampe und zumindest ein Reflektor dahinter angeordnet sind. Ferner ist im unteren Bereich des Gehäuses 3 eine Schaltuhr 4 zur Einstellung und Begrenzung der Bestrahlungszeit vorgesehen.

Bei der in Fig. 4 gezeigten, ersten Ausführungsform sind vier Lampen vorgesehen, und zwar jeweils zweiseitig gesockelte und im wesentlichen parallel zueinander angeordnete Leuchtstofflampen 5, 6, 7 und 8 mit einem Kolbendurchmesser von wahlweise 16 mm bis 38 mm und einer Länge von wahlweise 300 mm bis 600 mm. Zwei von ihnen emittieren im blauen Bereich und zwei im roten Bereich, wobei die Anordnung so getroffen ist, daß die Leuchtstofflampen 5 und 7 im blauen und die Leuchtstofflampen 6 und 8 im roten Bereich emittieren, derart, daß sich in der Lampenanordnung vorzugsweise jeweils eine blaue und eine rote Lampe abwechseln.

Bei der in Figur 5 gezeigten Ausführungsform sind in der Leuchte 2 zwei einseitig gesockelte Leuchtstofflampen 9 und 10 nach DIN EN 60901 mit Sockeln 2G11 und mit zwei durch einen Steg 11 miteinander verbundenen Schenkeln 12, 13 bzw. 14, 15 mit einer Gesamtlänge von wahlweise 225 mm bis 415 mm angeordnet, wobei jeweils der eine Schenkel 12 bzw. 14 einer Lampe 9 bzw. 10 im erfindungsgemäßen blauen Bereich und der andere Schenkel 13 bzw. 15 derselben im erfindungsgemäßen roten Bereich emittiert.

Beide Lampen 1 und 2 besitzen jeweils einen einzigen, gleichartig ausgebildeten Reflektor 16.

Patentansprüche

1. Verfahren zur kosmetischen Behandlung von Akne vulgaris durch Bestrahlung der betreffenden Partien mit Licht, **dadurch gekennzeichnet**, daß Licht mit zwei Emissionsspektren verwendet wird, von denen das eine im blauen Bereich (A) von 400 nm bis 450 nm und das andere im roten Bereich (B) von 580 nm bis 630 nm liegt.
2. Verfahren nach Anspruch 1, **dadurch gekennzeichnet**, daß die Bestrahlung einmal pro Tag für etwa fünfzehn Minuten durchgeführt wird.
3. Vorrichtung zur Ausführung des Verfahrens nach Anspruch 1 oder 2, **gekennzeichnet** durch eine Leuchte (1, 2) mit einem Gehäuse (3), in dem zumindest eine Lampe (5, 6, 7, 8; 9, 10) und zumindest ein Reflektor (16) dahinter angeordnet sind, sowie mit einer Schaltuhr (4) zur Einstellung und Begrenzung der Bestrahlungszeit.
4. Vorrichtung nach Anspruch 3, **dadurch gekennzeichnet**, daß in der Leuchte (1) vier zweiseitig gesockelte und im wesentlichen parallel zueinander angeordnete Leuchtstofflampen (5, 6, 7, 8) mit einem Kolbendurchmesser von 16 mm bis 38 mm und einer Länge von 300 mm bis 600 mm vorgesehen sind, von denen zwei (5, 7) im blauen Bereich und zwei (6, 8) im roten Bereich emittieren.
5. Vorrichtung nach Anspruch 4, **dadurch gekennzeichnet**, daß sich in der Anordnung innerhalb der Leuchte jeweils eine blaue (5, 7) und eine rote (6, 8) Lampe abwechseln.
6. Vorrichtung nach Anspruch 3, **dadurch gekennzeichnet**, daß in der Leuchte (2) zwei einseitig gesockelte Leuchtstofflampen (9, 10) nach DIN EN 60901 mit Sockeln 2G11 und mit zwei durch einen Steg (11) miteinander verbundenen Schenkeln (12, 13; 14, 15) mit einer Gesamtlänge von 225 mm bis 415 mm vorgesehen sind, wobei jeweils der eine Schenkel (12, 14) einer Lampe (9, 10) im blauen Bereich und der andere Schenkel (13, 15) derselben im roten Bereich emittiert.

Fig. 1

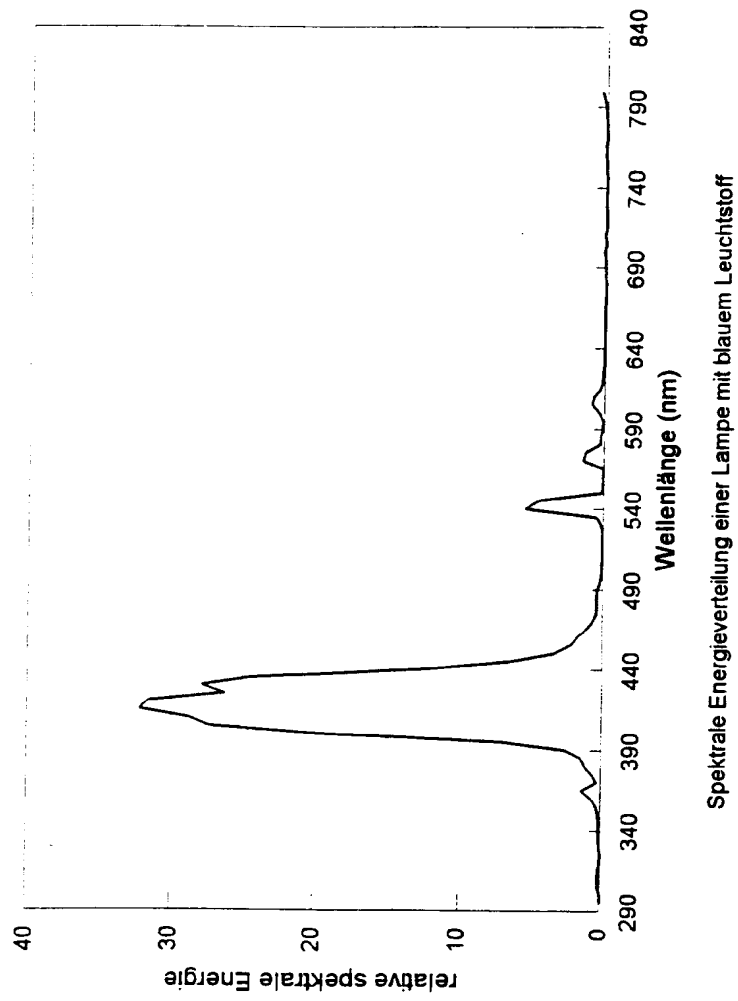
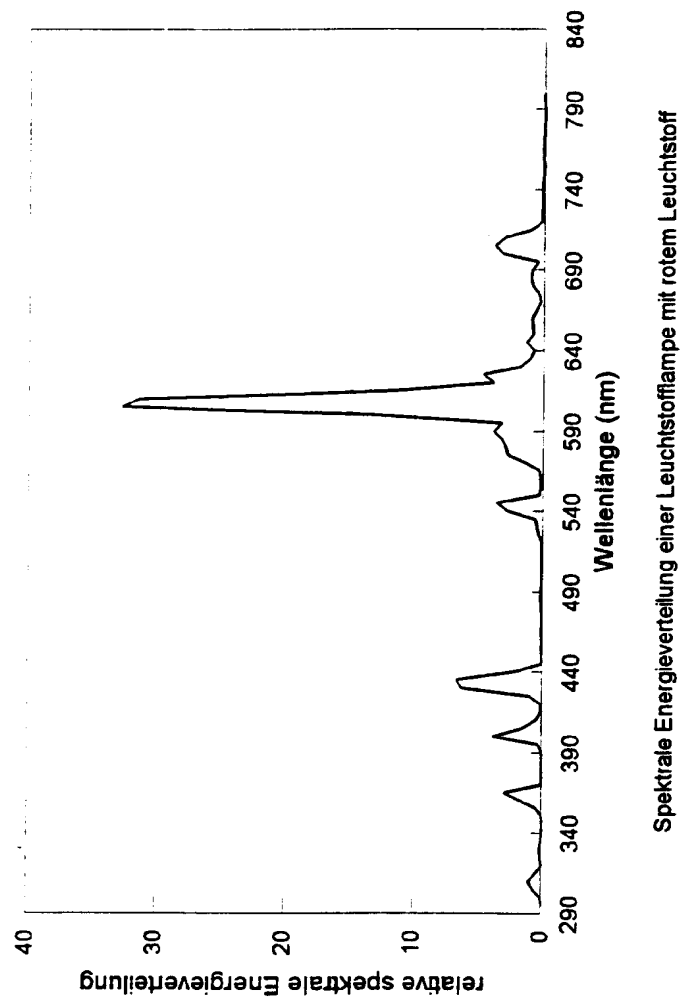


Fig. 2



Spektrale Energieverteilung einer Leuchtstofflampe mit rotem Leuchtstoff

Fig. 3

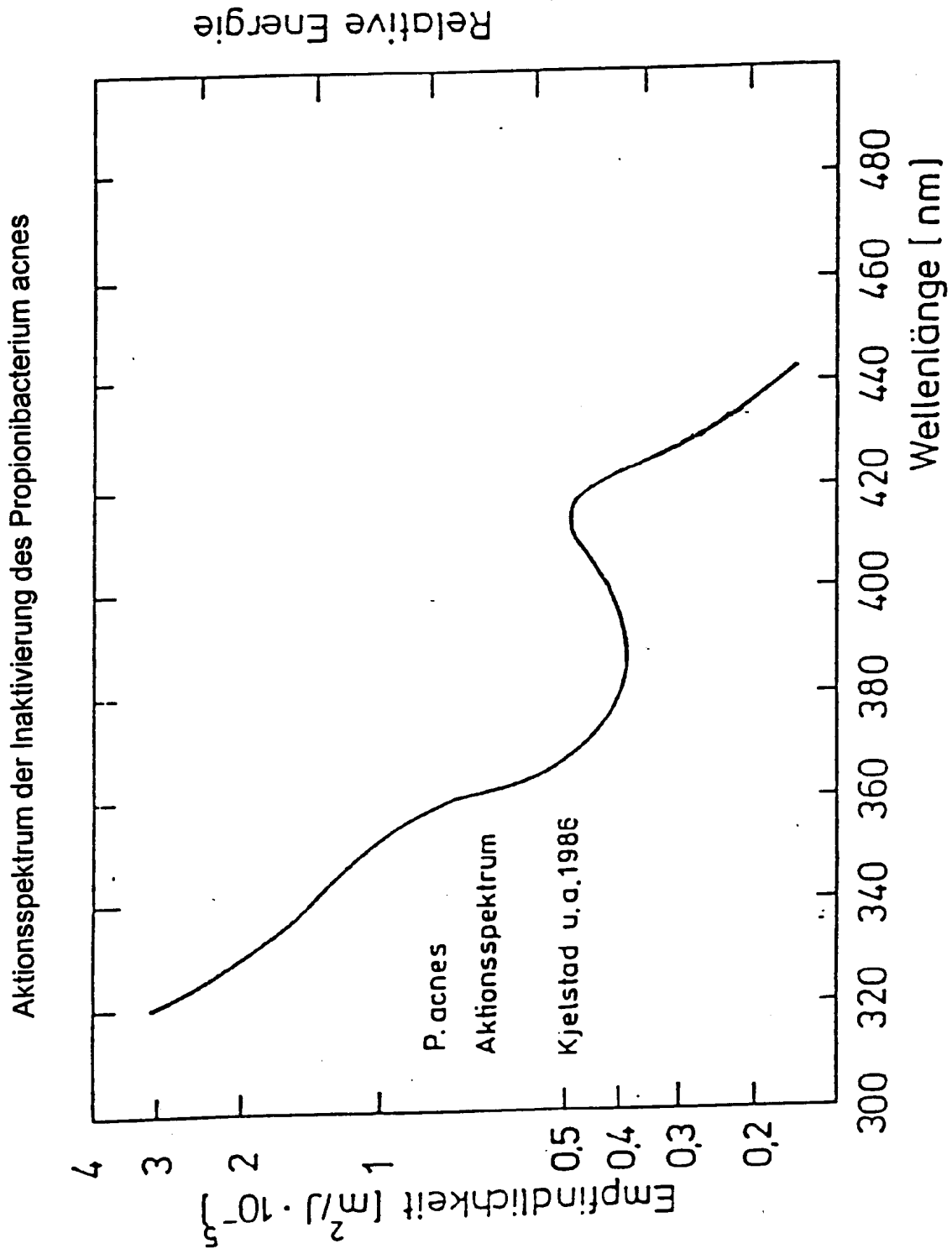


Fig. 4

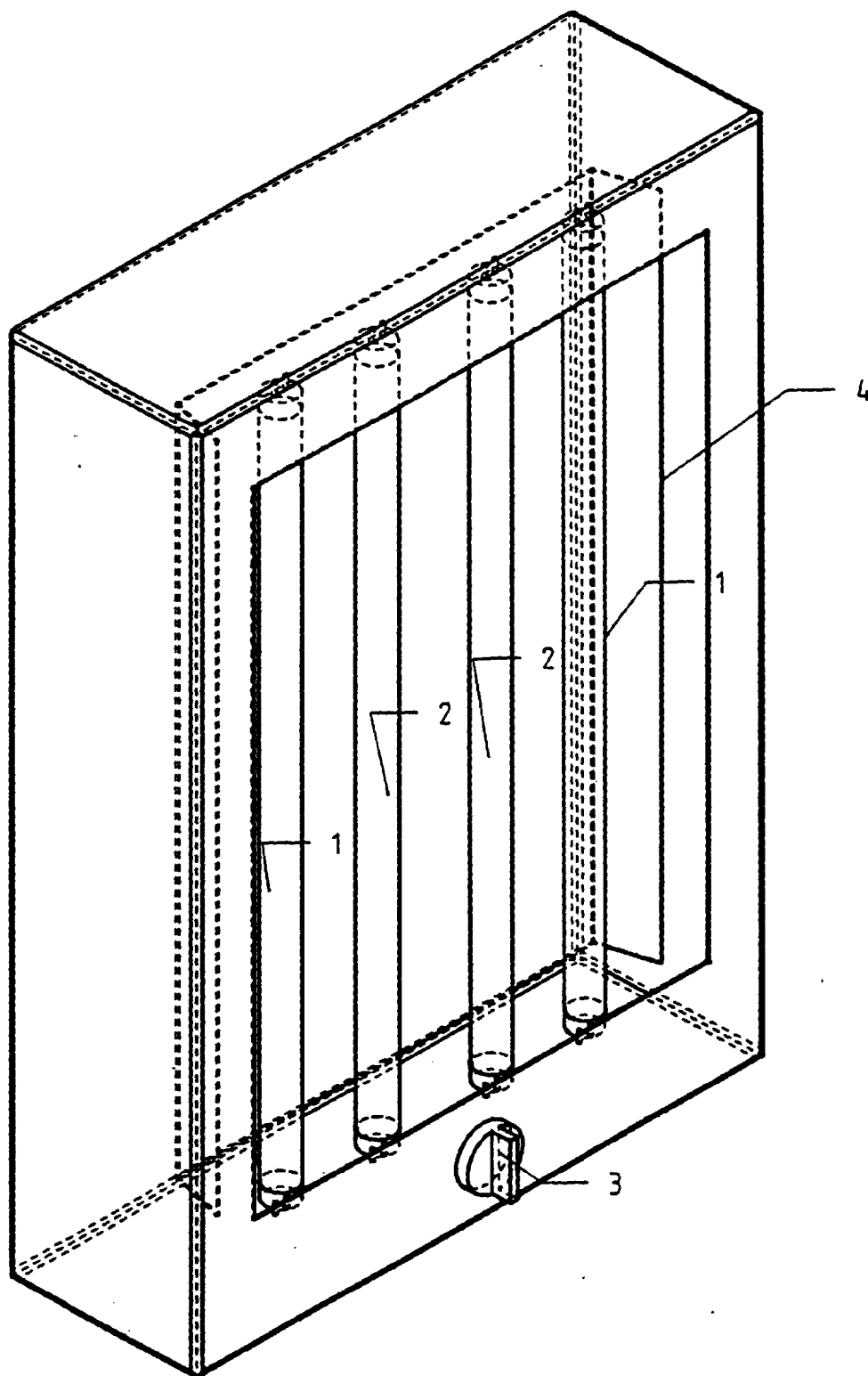
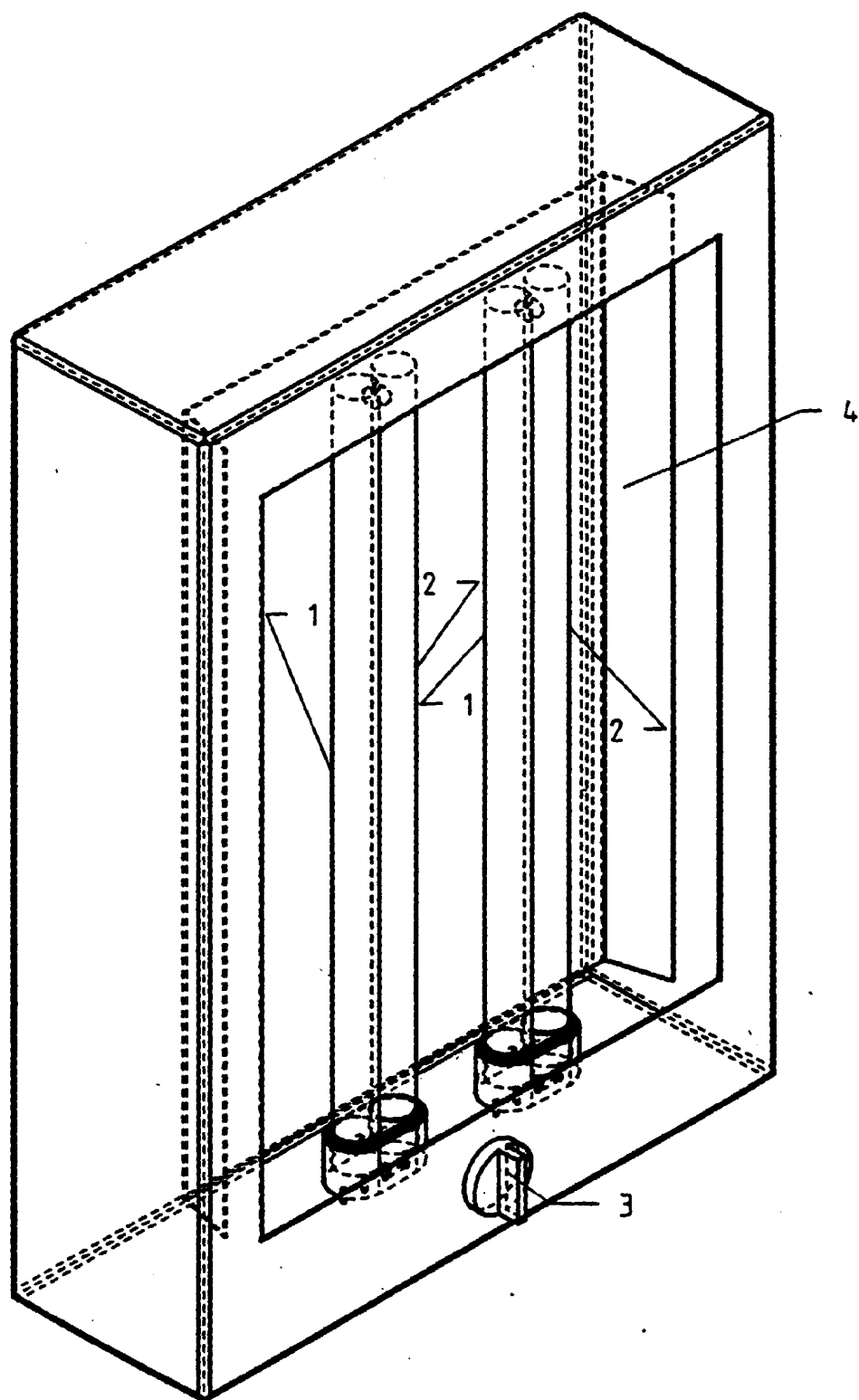
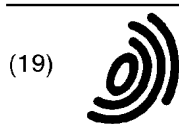


Fig. 5





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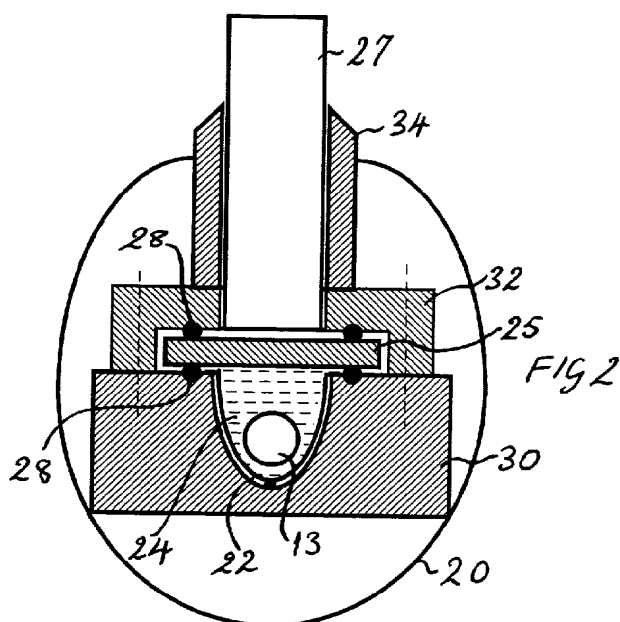
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(54) **Light pulse generating apparatus and cosmetic and therapeutic phototreatment**

(57) Apparatus for therapeutic or cosmetic phototreatment comprises a flash lamp (13) and a lamp operating circuit. The lamp is cooled by water in contact with the lamp which acts as an infra-red filter to reduce skin burning. Light from the lamp reaches the skin through a light guide (27) which has a convex curved end to focus the light and to press away haemoglobin in the target area or has a concave end to reduce pres-

sure on the skin depending on the desired treatment. Relatively long and low power square shaped power pulses drive the lamp to produce light output pulses adapted to the relaxation time of the target structure to maximise the heating of the target whilst minimising heating of the skin surface. Target structures may be blood vessels or hair follicles. Automatic detection of a glass filter (25) may be provided.



EP 0 885 629 A2

Description

The present invention relates to apparatus for producing light pulses and to apparatus and methods using light for cosmetic or therapeutic photo-treatment.

Apparatus of this type may be used for therapeutic purposes including treatment for psoriasis, vascular traumas, telangiectasis, capillary hemangioma, cancerous cells, portwine stains, and birthmarks and also for depilation.

WO 091/15264 describes a device for treatment of undesired skin disfigurements, which apparatus comprises a gas filled lamp and an optical filter. The lamp is powered by a capacitor charged to about 2,000 volts, which is sufficient for generating a pulse with a pulse duration between $\frac{1}{2}$ and 1 ms.

EP 0 565 331 A2 describes a therapeutic treatment device which includes a gas filled flash lamp and a set of optical filters. The power circuit includes three different pulse forming networks, which may be triggered selectively or successively. Each pulse forming network includes a capacitance and an inductance and a relay contact to trigger the discharge. The three networks are designed for providing different pulse widths. The capacitors are charged to a voltage, typically in the range of from 500 volts to 5 kilovolts.

EP 0 736 308 A2 discloses a method and an apparatus for depilation, wherein an apparatus of a similar kind is used. According to this publication an energy fluence in the order of from 10 to 100 J/cm² is used for the purpose of killing hair follicles without burning the surrounding skin. The publication also suggests the use of an optically transparent water-based gel applied to the skin prior to treatment and serving the purpose of providing a heat sink to prevent hyperthermia in the skin.

We have found several limitations and problems in devices and methods according to the prior art. Damage to the skin or burns occur very easily, making it difficult to find an energy level at which the pulses will be effective but will not harm the skin. Further the amount of heat absorbed in the optical filters may be excessive, frequently leading to destruction of the filters.

These problems are addressed by the various aspects of the invention set out below which may be used separately or in any combination.

In a first aspect, the invention provides apparatus for producing a pulse of light, comprising a light source operable to produce a said pulse of light, and a filter for filtering undesired light output frequencies from said pulse, characterised in that said filter comprises water.

The water acts as an ideal infra-red filter for apparatus to be used in treating the skin, because it is mainly the water content of the skin which adsorbs infra-red energy to produce undesirable generalised heating in phototreatments.

Preferably, the apparatus comprises means for defining a flow path for said water, which means is optically transparent at least in a region in which said water acts

as said filter, and means for producing a flow of said water through said flow path.

The light source preferably forms part of the means defining said flow path for water, whereby said water acts both to filter said light pulse and to cool said light source.

The distance travelled through the water by the light rays on average is preferably at least 1 cm, more preferably at least 1.5 cm. It may be selected to achieve 50% attenuation of the output at 1200 nm of the lamp.

This aspect of the invention includes the use of water as an infra-red absorbing filter in apparatus for the cosmetic or therapeutic photo-treatment of the human or animal body.

Photo-treatment apparatus has been described previously, e.g. in EP-A-0565331, which features a light guide to be placed against the skin. Non-contacting light guides terminating in focussing lenses are also known for use in photo-treatment apparatus, e.g. in US-A-3327712. The light guide in EP-A-0565331 has a flat distal end for making contact with the skin. We have now discovered that it is advantageous to provide a convex curved distal end in such a light guide. Pressing the curved surface of the light guide against the skin in depilation treatment one or two millimeters into the tissue leads to removal of haemoglobin and oxyhaemoglobin from the light path towards a hair follicle. Haemoglobin and oxyhaemoglobin absorb the energy in the bandwidths used for depilation. By gently pressing the haemoglobin and oxyhaemoglobin away with the curved tip of the light guide, more energy is delivered to the target chromophore (melanin at the hair bulb) with a lower energy output than previously (i.e. there is an improved ratio between useful and damaging heat absorptions). Furthermore, the curved shape allows the light guide to slide easily over the skin when covering treatment areas larger than the "foot print" of the light guide.

In a variant of the first aspect of the invention there is provided an apparatus for producing a pulse of light, comprising a light source operable to produce a said pulse of light, and a filter for filtering undesired light output frequencies from said pulse, characterised in that said filter comprises a liquid within a conduit and the apparatus further comprises means defining a flow path for said liquid, a port of said flow path being constituted by said conduit and means for passing said liquid through said flow path.

Accordingly, in a second aspect, the invention provides photo-treatment apparatus comprising a light source and means for transmitting light output from the light source to a treatment site, said means including a light guide having a distal end for contacting the skin of a patient for said photo-treatment, said light guide distal end being shaped in a convex curve whereby pressing the light guide gently against the skin of the patient reduces the amount of blood in the skin below the light guide.

Preferably, said light guide is shaped as a paralleli-

pedic prism with a bull-nosed projection on said distal end. The projection is preferably of uniform transverse cross-section.

Pressure should not be used by the operator when treating vascular lesions because the haemoglobin and oxyhaemoglobin are then the target.

Indeed, for use in such treatment, the invention provides, in a variant of its second aspect, photo-treatment apparatus comprising a light source and means for transmitting light output from the light source to a treatment site, said means including a light guide having a distal end for contacting the skin of a patient for said photo-treatment, said light guide distal end being shaped in a concave manner whereby to relieve pressure applied to the skin by the light guide in regions where blood is a target of said light output.

According to the prior art, the power supply typically comprises a capacitor which is discharged to feed current through a series inductance and into the arc lamp. The prior art circuit creates a current pulse shaped approximately as one half period of a hyperbolic sine wave, rising from zero to a maximum and declining again to zero. We have found that the relatively gradual rise rate in the light power output at the commencement of each pulse is disadvantageous, as further described below.

In a third aspect, the present invention provides apparatus for producing a pulse of light, comprising:

a light source operable to provide an output of light in response to a power input, and

a power supply connected to the light source for providing said power input,

wherein said power supply is operable to provide a power output pulse or pulse train to drive said light source, to produce a light output pulse or pulse train during which light output pulse or pulse train for at least 80% of the light output period the light power output is from 75 to 125% of the time-weighted average light power output during the light output period.

This implies a much more rapid rate of rise in power at the start of the pulse and a higher rate of fall in power at the end of the pulse than is provided by a hyperbolic sine wave pulse power profile. This aspect of the invention aims at driving the light source on a current pulse which typically is shaped approximately as a square wave. Although the power requirement is indicated above in terms of the light power output, it will be understood that this is generally dependent on there being an electrical power input into the lamp (i.e. the output of the power supply) satisfying similar requirements in terms of electrical power.

Preferably, for at least 90% of the light pulse light output period the power output is from 75% to 125%, more preferably 90 to 110%, of the time-weighted average power output during the pulse light output period, most preferably from 95 to 105% of said value. Alterna-

tively, it is preferred that for at least 95% of the light output period the light power output is from 75 to 125% of the time-weighted average light power output during the light output period, more preferably from 90 to 110% of said average value and most preferably from 95 to 105% of said value.

The time weighted average power output is given by the formula:

$$\sum_t p_n \cdot t_n$$

where p_n is the power p at time t_n and t is the total pulse light output period Σt_n .

Alternatively, the time from the commencement of the light pulse to the light pulse reaching full power is no more than 10 percent, more preferably no more than 1 percent of the light output period of the pulse.

Preferably, means is provided for adjusting said time-weighted average light power output.

Arc lamps may be driven at different current levels, outputting different levels of optical radiation. However, this causes an associated shift in the spectrum. Thus the variation of the radiation in the short wave length range, e.g. below 700 nm tends to vary more than the variation in the longer wave length range. For instance on recording the optical output spectral density generated while driving an arc lamp at 100% and at 50% of maximum rated current, it was found that the optical output spectral density at 900 nm dropped to 67% of the preceding value, whereas the output at about 480 nm dropped to about 40% of the respective preceding value.

The unfiltered optical radiation from a flash light source is mainly from 380 nm to 1250 nm. For depilation, optical radiation at wave lengths longer than 700 nm is favored for its ability to penetrate deeper into the skin. Thus, for this application the short wave length radiation is regarded as an unwanted output to be filtered away in optical filters. The result is an excessive power dissipation in the filters.

The changing spectrum of the arc lamp if driven on a varying current as provided by power supply units according to the prior art, also makes it very difficult to estimate the effective output after the filters. According to this aspect of the invention the lamp is driven on a substantially constant level of current, where the spectrum then is approximately unchanged and the energy delivered to the target is proportionate to the duration of the square wave pulse or pulses.

The generation of this kind of drive signal requires a different electric circuit, which operates with a higher intrinsic power loss compared to the circuits of the prior art. The invention therefore achieves the result referred to only at the cost of an increased power loss in the electric driving circuit. The shifting of the power loss to take

place in the power supply unit rather in the optical filters is considered a major advantage, since power dissipated in the power supply unit may easily be kept away from the patient and from the sensitive optical components in the applicator.

The capability of the apparatus according to the invention of reducing the amplitude of the power output may be used to a substantial advantage. We have found that the skin epidermis is capable of dissipating a greater amount of energy input than the tissue in the hair follicle. The difference, which may be in the order of factor of 20, is attributed to the greater thermal conductivity in the epidermal region. This difference more than offsets the disadvantage of the lowered intensity at the hair follicles.

Hence, for a given irradiation input, the hair follicle is heated to a higher temperature than the epidermis. This means that it is possible to establish a level where the hair follicles may be killed without harming the epidermis. The exact level may vary depending on skin pigmentation and has to be established for the particular patient. The procedure of establishing a proper level is considered to lie within the capabilities of those skilled in the art. The apparatus according to the invention is capable of outputting a treatment signal at a power level which is accurately controlled, in order to provide just the desired irradiation input.

Preferably, a single light pulse is applied to each treatment location.

The duration of each such pulse may be up to 100 msec, e.g. from 2 to 100 msec, more preferably from 10 to 70 msec. Preferably, the pulses are relatively long and relatively low in power rather than delivering the same amount of energy to the treatment site by being higher in power and shorter. The latter is what is obtained using laser systems. In the case of depilation and vascular treatments, this allows advantage to be taken of the differing thermal relaxation times of the skin and the hair follicle, so that heat is permitted to be lost from the skin, preventing burning, while heat accumulates in the hair follicle or blood vessel. Thus, the optimum pulse duration can be determined according to a relaxation time based algorithm.

By way of further guidance, it is generally accepted that the relaxation time of a hair follicle varies according to the thickness of the individual hair such that it is about 15ms for a thin hair, about 50ms for thick hair such as pubic hair, and for normal thickness hair about 30ms. It is preferable that the pulse duration is approximately equal to the relaxation time of the target hair follicle or other target structure. The relaxation time of the skin surface is only about 2ms. Accordingly, even relatively short pulses (e. g. 5ms) will heat the skin surface to about the temperature it would reach if the application of energy was continuous. However, because of the longer relaxation time of the hair follicles, these will not be heated to close to their equilibrium value by such short pulses. Accordingly, where a very short pulse time

is used, it will be necessary to use a much higher light intensity per unit area in order to heat the hair follicles adequately, with as a consequence greater heating of the skin surface. When a relaxation time algorithm dictated pulse duration is used, the hair follicle is heated to a temperature closer to the maximum temperature it would reach under continuous illumination at the light intensity used, which may therefore be much reduced compared to that needed with a short pulse duration. The skin surface is better able to lose heat and accordingly remains acceptably cool whilst the hair follicle or other target structure is heated to a higher temperature, sufficient to cause the desired amount of damage to the target cells.

According to any of the various aspects of the invention the light source preferably comprises a gas-filled arc lamp. Preferably, said gas-filled arc lamp is a xenon or krypton lamp.

However, the use of long pulses (e.g. 15 msec. or more) may shorten the life of gas discharge lamps and accordingly, we prefer to divide each long pulse into a pulse train of closely spaced shorter pulses, each having the power profile specified above. The duration of each such shorter pulse may be from 2 to 25 msec and the spacing between such shorter pulses is preferably from up to 3 msec, e.g. 0.2 to 1.5 msec.

The total light output period of a pulse train is equal to the sum of the durations of the small pulses within the pulse train. This will differ from the duration of the pulse train because it omits the intervals between the pulses in the pulse train. Thus for a pulse train of χ pulses, each of duration t_{χ} and each spaced from the succeeding pulse by an interval i_{χ} the period of light output D of the pulse train is given by:

$$D = \Sigma t_{\chi}$$

and is related to the duration T of the pulse train by:

$$D = T - \Sigma i_{\chi}$$

For a single long pulse, the light output period of course equals the duration as there are no intervals and $\Sigma i_{\chi} = 0$.

The total light output period of the pulse train is related to the duration of the pulse train also by the duty cycle which is the fraction of the pulse train duration during which there is light output.

Because the use of longer pulse or pulse train durations allows a reduced maximum light intensity to be employed, in another aspect, the present invention provides an apparatus for producing a pulse of light, comprising a light source operable to provide an output of light in response to a power input, and a power supply connected to the light source for providing said power input, wherein said power supply is operable to provide

a power output pulse or pulse train to drive said light source to produce said light output pulse or pulse train, such that the ratio of the maximum light intensity to the total light energy output is no more than 75 kw : 1 kw sec, preferably 50 kw : 1 kw sec.

The apparatus preferably further comprises a housing for said light source, an aperture defined by said housing and a reflector in said housing positioned to reflect a beam of light through said aperture.

The apparatus preferably further comprises an optical filter in the path of said light beam, said optical filter being adapted to pass only selected wavelengths of said light, preferably this is water.

If apparatus for photo-treatment is to be used for treating a variety of conditions, it will normally be necessary that the light output be filtered differently according to the condition treated. There will normally therefore be provided a plurality of filters having differing filter characteristics and it will often be a matter involving the safe operation of the apparatus that the filter or the appropriate one be in place before treatment commences.

Accordingly there is provided according to a further aspect of the invention apparatus for photo-treatment comprising a light source, means for receiving a filter adapted to pass only selected wavelengths of light so as to dispose said filter in a light path from said light source, sensor means for detecting the presence and/or nature of a said filter in said filter receiving means, and interlock means for preventing operation of said light source to carry out photo-treatment except when a said filter or a said filter appropriate to an intended photo-treatment is present in said receiving means and/or for providing an alarm signal if a said filter or appropriate filter is not present in said receiving means.

This may be achieved by placing an electrically resistive circuit path, e.g. a resistive track, on the filter and providing means in the filter receiving means for measuring the electrical resistance of the track. Different filters will be provided with different resistances.

Alternatively, this may be achieved by a mechanical sensor detecting the presence of a filter in the filter receiving means, e.g. a sprung plunger which is depressed by the introduction of a filter and which actuates an electrical switch to make or break a circuit to allow operation of the apparatus. Different filters may be mechanically sensed by respective switches if they differ physically in thickness or shape. Preferably however there is at least one light source and at least one light sensor operatively arranged to observe the filtering activity of the filter on transmitted or reflected light and suitable electronic control means to determine the presence and optionally the nature of the filter from the light detected by said sensor or sensors.

The filters employed in photo-treatment apparatus of this type in the past have been of the interference type which comprise a non-filtering substrate bearing an extremely thin coating, typically of several layers of the order of the wavelength of light in thickness. We have

found that at the high light intensities needed in photo-treatment apparatus, such filters are prone to damage. Any particle of dust on the filter surface can adsorb the light energy and become heated to a temperature which causes a pin-hole to form in the filter coating, allowing unfiltered light through.

The defect will progressively increase in size with further use and will eventually destroy the efficacy of the filter, producing progressively more severe burning of the skin.

Accordingly, there is provided according to a further aspect of the invention apparatus for photo-treatment comprising a light source, a filter adapted to pass only selected wavelengths of light disposed in a light path from said light source, said light source being adapted to produce a light flux of at least 250 J/cm²/sec, wherein said filter is a non-interference absorption filter.

Such a filter is preferably of heat resistant glass or other heat resistant transparent material having a pigment distributed throughout its thickness. Additionally, a coated reflection filter may be formed on the surface of such a filter nearer the light source to reflect energy away so as to reduce adsorption in the filter and heating of the filter.

It is known to cool the lamp in this type of apparatus by circulating a cooling liquid, normally water, through channels in the material of the lamp housing. We have appreciated that this is a less than ideal approach. Accordingly, we prefer to provide means for circulating a cooling liquid in contact with said light source.

The cooling liquid may also circulate in contact with one or more filters for the light output. Thus, preferably, such apparatus further comprises a filter adapted to pass only selected wavelengths of light disposed in a light path from said light source and also in contact with said cooling liquid.

The cooling liquid is preferably water. Not only does water have a direct cooling function in the apparatus itself, we have appreciated that it acts as an ideal infra red filter, being adapted to filter out just those wavelengths which would cause unwanted heating of the skin of the patient, in which water is a principal infra red absorber. More generally as indicated in relation to the first aspect of the invention, the invention includes the use of water in the light path of photo-treatment apparatus as an infra red filter. Whilst this is preferably achieved by circulating the water over the lamp as a coolant as described above, water instead may be contained in a vessel in the light path to act purely as a filter. By the term 'water' in this context we include not only pure water but aqueous liquids generally, and in the case of the filter described above we include non-flowable transparent aqueous materials such as gels also.

As the coolant liquid may expand due to heating, the apparatus preferably includes a pressure relief device.

The power supply for driving the light source preferably comprises a capacitor, a charging circuit adapted

for charging the capacitor to a preselected voltage, a resistor in series between said capacitor and said light source and a discharge switch operable to change from a non-conductive state to a conductive state to cause said capacitor to discharge said light source and back to said non-conductive state again.

Preferably, when the light source is an arc lamp the power supply comprises a simmer generator adapted for feeding the arc lamp with power at a level which is sufficient to keep the arc in the conductive state.

Optionally, the apparatus may include a light measuring device for measuring light reflected from or transmitted through the skin. The source of the light may be the main light source of the apparatus or may include a respective light source providing light to be reflected by the tissue. The measurement of the reflection of particular wavelengths or wavelength bands of light may be used to form a judgement regarding the skin coloration which in turn may be used in calculating the appropriate photo-treatment.

The invention further provides a method of treating live tissue for cosmetic purposes using apparatus according to any aspect of the invention. Cosmetic treatments envisaged include hair depilation, tattoo removal, wrinkle smoothing, skin rejuvenation, removal of disfiguring skin ailments and birthmarks.

The invention further provides a method for treating live tissue for therapeutic purposes using apparatus according to any aspect of the invention. Therapeutic purposes envisaged comprise treatment for psoriasis, vascular traumas, telangiectasis, capillary hemangioma, cancerous cells, removal of birthmarks, etc. Such treatment methods include not only those in which the light acts directly on the tissues but also methods of photodynamic therapy in which the light acts on a substance applied to or administered to the body and activates the substance, e.g. cleaves a prodrug to release the active material at a treatment site or causes the production of singlet excited oxygen at said treatment site as described in "Chemistry in Britain" May 1998 pp 45 - 50.

Further objects, features and advantages of the invention will appear from the appended detailed description given with reference to the enclosed drawings, wherein

Figure 1 illustrates a circuit diagram of a power supply with a lamp according to the invention;

Figure 2 illustrates a transverse cross-section through a photo-treatment apparatus according to the invention including for use with the power supply arrangement of Figure 1;

Figure 3 is a longitudinal cross-section through the apparatus of Figure 2;

Figure 4 is a time chart of the current fed through the lamp of Figure 1 during a pulse;

Figure 5 is a chart of the luminous spectral density of the output of a Xenon lamp driven according to the prior art and the circuit of Figure 1; and

Figure 6 shows three pulse trains suitable for depilation treatment.

Figure 7 shows a plot of light output power against time produced by machine settings for thin hair;

Figure 8 shows a similar graph for machine settings for thick hairs;

Figure 9 shows a plot of light absorbance by different distances travelled in water;

Figure 10 shows a light guide for use in accordance with the second aspect of the invention and as an aid for focussing the light output to a distance below the skin in apparatus as shown in Figures 1 to 3; and

Figure 11 shows a second form of light guide suitable for use when treating blood vessel disorders.

All drawings are schematic, illustrating only the principal features of the apparatus, including those essential to enable those skilled in the art to practise the invention whereas other features are omitted from the drawings for the sake of clarity. Throughout the drawings the same references are used to designate identical or similar features.

Reference is first made to Figure 1 illustrating the driving circuit and the lamp.

The circuit in Figure 1 comprises a control unit in the form of a personal computer 10 (PC) used to control the system. The PC is connected to a power supply 12 which is adapted to charge through the diode D1 the capacitor C. In a preferred embodiment the power supply is adapted to charge the capacitor to a voltage set from the PC to a level in the range from 100 - 1000 volts. In a preferred embodiment the capacitance of the capacitor C is 10 to 100 mF.

The capacitor C is connected through diode D2 and resistor R to the flash lamp 13. The circuit is completed by a solid state switch IGBT, which is in the preferred embodiment implemented in the form of an isolated gated bi-polar transistor. The IGBT is controlled from the PC by a line 14. The IGBT is capable of changing from non-conductive to conductive state, of carrying currents in the range of 500 A and of changing from conductive to non-conductive state again, breaking this current.

On the right hand side of Figure 1 a simmer power supply 16 is illustrated. This power supply is capable of feeding the flash lamp through the diode D3 with a simmer current in the order of 100 mA. In order to ignite the flash lamp the simmer power supply outputs a short pulse at a voltage of about 10 - 20 kilovolts on an electrode 18. The simmer current maintains a narrow arc

inside the gas-filled lamp to keep this lamp in the conductive state.

Reference is now made to Figure 2 illustrating a vertical transverse section through an applicator according to the invention. The applicator comprises housing 20, lamp 13, and a reflector 22 surrounding the lamp, long wave pass filter 25, light guide 27 and pressure relief o-rings 28. The reflector 22 is of ceramic and lines a U-shaped channel in a block of stainless steel 30. A second block of stainless steel 32 is bolted over the reflector. The filter 25 is sandwiched between blocks 30, 32 and is spaced from each of them by o-ring seals 28. The light output guide 27 is received in an opening in the block 32 and is supported by a rectangular section collar 34. The reflector 22 is shaped to direct the light output of the lamp upwards as illustrated in the figure. The edge of the reflector 22 constitutes a light output aperture.

The reflector together with the filter 25 forms a chamber 24 which is filled with water. As shown in Figure 3, the block 30 has a tubular cavity in which the lamp 13 is received and is sealingly supported by o-rings 36. A water inlet 38 and a water outlet 40 communicate with respective annular spaces surrounding the ends of the lamp 13, which communicate also with and form part of the chamber 24.

An alternative form of pressure relief device (not shown) comprises a bulb-like expanded chamber filled with air and in fluid communication with the chamber 24. This air-filled chamber acts like a spring capable of smoothing out any pressure shocks in the water chamber that may be caused by the sudden discharges of the lamp.

The long wave pass filter 25 is adapted to absorb a part of the light in the near UV range e.g. UV and near UV shorter than 510 nm. It is a heat resistant glass filter of the non-interference type. IR light is absorbed by the water.

Optionally an additional removable filter may be provided between the proximal end of the light guide 27 and the filter 25. This may be removable and changeable by the user.

One may thus pass light of wavelength from 510 to 600 nm, e.g. 510 to 590 nm, according to the therapeutic requirement.

The removable filter is then chosen according to the intended treatment and is of coloured heat resistant glass (optionally combined with coated reflection filters and coloured filters) and may be substituted with other filters of similar type in order that the operator may choose from a selection of filters with different optical band widths. It may be a band pass filter.

Reference is now made to Figure 4 illustrating a time chart of the current fed through the Xenon lamp during a pulse of treatments. Figure 4 illustrates a pulse of a duration of 100 ms. The pulse rises practically immediately to a level of 338 A and decays exponentially to about 276 A after 100 ms.

Thus, the circuit of the preferred embodiment ap-

proximates the desired square wave by a sloping exponential decay with a time constant depending on the capacitance of the capacitor, the series resistor R, and the current driven through the Xenon lamp. Generally, satisfactory results are achieved if the current decays from 100% to somewhere above 50% of the initial value.

Reference is now made to Figure 5 illustrating a chart of power spectral density of the radiation output by the Xenon lamp. Figure 5 comprises two graphs, one drawn for a Xenon lamp at a current density of 2400 A per cm² and illustrating the optical output from a wave length about 420 nm up to about 1100 nm, the other curve showing the output of a current at half of this level. The curve illustrates the fact that the spectral output drops in a frequency dependent manner on the reduction of the drive current. For instance at 900 nm the output drops to approximately 65% while the output at 480 nm drops to about 40%, both taken relative to the respective preceding values.

Figure 6 shows three different pulse trains for electrical power input suitable for use as depilation treatment using the apparatus of the invention. Pulse train (a) consists of five pulses of 3 ms duration spaced by delay intervals of 1.5 ms. The light output period of the pulse train is therefore 15 ms. The time weighted current average of the pulse train is 280 A and the maximum and minimum values are 300 A and 250 A respectively. Thus for essentially 100% of the light output period, the power will be within the 75% to 125% of average band. This is an example of a pulse train suitable for use in depilation of thin hairs.

The pulse train in (b) comprises four pulses of 5 ms with an interval of 1.5 ms. The average, maximum and minimum currents are 250, 285 and 200 A and again this is within the 75% to 125% of average band for essentially 100% of the light output period. This is suitable for depilation of normal hairs.

In (c) the average, maximum and minimum currents are 150, 190 and 95 A and the power is within the 75% to 125% band for approximately 80% of the light output period. The illustrated pulse train is suitable for depilation of thick hairs.

For each individual short pulse in each pulse train illustrated, the requirement for a power within the 75% to 125% of average power is met for at least 90% of the light output period.

Figure 7 shows a plot of light output power in kW against time in seconds for a typical apparatus according to the invention (solid lines) compared to a commercially available machine (dotted lines), both being adjusted for thin hair. It will be seen that by comparison, the light output period of the apparatus of the invention is longer and the maximum power lower, although the energy output (area under each graph) is the same. Figure 8 shows the same but with the adjustment set for thick hairs. The effect is even more pronounced. The ratio of peak output power to energy supplied in Figure 7 is about 50 sec⁻¹ for the apparatus according to the

invention, but about 130 sec^{-1} for the conventional machine, whereas in Figure 8 the ratios are about 30 sec^{-1} and 130 sec^{-1} , respectively.

In order to use the system for treatment, an operator would place the applicator adjacent a selected treatment area and set the control unit to carry out an initialization routine. As part of this routine, the operator would enter data into the control unit concerning the patient and the type of treatment desired. Optionally, the control unit is programmed to ignite the flash lamp and burn it on the simmer power supply in order provide a low level of irradiation, by which the control unit through the utilization of a photodetector will establish the reflectivity or transmission value of the treatment area. These data enable the control unit to suggest an appropriate irradiation scheme, which may comprise pulse level and pulse duration.

Once the operator has accepted the treatment scheme, he will only need to move the applicator to the respective treatment areas and activate a flash trigger, while the control unit will verify that contact is established, and that the reflectivity has the presumed value, and will then output the appropriate treatment signal.

As shown in Figure 10, the apparatus may include a light guide with a curved distal end. The illustrated light guide is a parallelepipedic prism 60 of rectangular transverse cross-section which has at its distal end a bull-nosed projection 62, such that in side view (on its smaller side face) the light guide is as shown in Figure 10 and is of constant cross-section. The convex nose of the light guide can be pressed against the skin 64 to reduce (oxy) haemoglobin absorption of the light as described above by driving blood out of capillaries 66 and compressing larger vessels 68 to reduce blood flow. The convex curve may also serve as means for focussing the light output to concentrate it at a selected level below the skin. This reduces the energy density at the skin surface for a given energy density at the treatment site. The focussing depth may be made adjustable by a focussing mechanism or by the provision of separate lenses that may be swapped or supplemented with one another.

As shown in Figure 11, instead of the light guide 11 having a convex nose to compress surface blood vessels, the light guide may on the contrary have a concave nose in order to minimise the pressure applied to surface blood vessels by contact with the light guide. This will be desirable when the blood vessels are themselves the target of the treatment.

Although various components, systems and methods have been explained in particular settings above, this is not to exclude that such components, systems or methods might be applied in other settings or applied differently. The particular examples mentioned have only been mentioned with the purpose of facilitating the understanding of the invention and not with the purpose of limiting the scope whereof which is defined exclusively by the appended patent claims.

Claims

1. Apparatus for producing a pulse of light, comprising a light source (13) operable to produce a said pulse of light, and a filter for filtering undesired light output frequencies from said pulse, characterised in that said filter comprises water.
2. Apparatus as claimed in Claim 1, comprising means (38,24,40) for defining a flow path for said water, which means is optically transparent at least in a region in which said water acts as said filter, and means for producing a flow of said water through said flow path.
3. Apparatus as claimed in Claim 2, wherein said light source forms part of the means defining said flow path for water, whereby said water acts both to filter said light pulse and to cool said light source.
4. Apparatus as claimed in any one of Claims 1 to 3, which is for use in a method of cosmetic or therapeutic photo-treatment of the human or animal body.
5. Apparatus as claimed in any preceding claim, wherein said light pulse has an energy of not less than $250 \text{ J/cm}^2/\text{sec}$.
6. The use of water as an infra-red absorbing filter in apparatus for the cosmetic or therapeutic photo-treatment of the human or animal body.
7. Apparatus for producing a pulse of light, comprising a light source (13) operable to produce a said pulse of light, and a filter for filtering undesired light output frequencies from said pulse, characterised in that said filter comprises a liquid within a conduit (24) and the apparatus further comprises means (38,24,40) defining a flow path for said liquid, a part of said flow path being constituted by said conduit, and means for passing said liquid through said flow path.
8. Apparatus as claimed in Claim 7, wherein said flow path forms a closed circuit around which said liquid circulates.
9. Apparatus as claimed in Claim 7 or Claim 8, wherein said liquid acts as an infra red filter
10. Photo-treatment apparatus comprising a light source (13) and means for transmitting light output from the light source to a treatment site, said means including a light guide (27,60) having a distal end (62) for contacting the skin of a patient for said photo-treatment, said light guide distal end being shaped in a convex curve whereby pressing the

light guide gently against the skin of the patient reduces the amount of blood in the skin below the light guide.

11. Apparatus as claimed in Claim 10, wherein said light guide is shaped as a parallelepipedic prism with a bull-nosed projection on said distal end.
12. Photo-treatment apparatus comprising a light source and means for transmitting light output from the light source to a treatment site, said means including a light guide having a distal end for contacting the skin of a patient for said photo-treatment, said light guide distal end being shaped in a concave manner whereby to relieve pressure applied to the skin by the light guide in regions where blood is a target of said light output.
13. Apparatus for producing a pulse of light, comprising: a light source (13) operable to provide an output of light in response to a power input, and a power supply (12) connected to the light source for providing said power input, wherein said power supply is operable to provide a power output pulse or pulse train to drive said light source to produce said light output pulse or pulse train, during which light output pulse or pulse train for at least 80% of the light output period (i.e. the duration of a single pulse or the aggregate of the duration of the pulses within a pulse train excluding intervals between pulses) the light power output is from 75 to 125% of the time-weighted average light power output during the light output period.
14. Apparatus as claimed in Claim 13, wherein for at least 90% of the light output period the light power output is from 75 to 125% of the time-weighted average light power output during the light output period.
15. Apparatus as claimed in Claim 13 or 14, wherein means is provided for adjusting said time-weighted average light power output.
16. Apparatus as claimed in any one of Claims 13 to 15, further comprising a housing (20) for said light source, an aperture defined by said housing and a reflector (22) in said housing positioned to reflect a beam of light through said aperture.
17. Apparatus as claimed in any one of Claims 13 to 16, further comprising an optical filter (25) in the path of said light beam, said optical filter being adapted to pass only selected wavelengths of said light.
18. Apparatus for photo-treatment comprising a light source (13), means for receiving a filter (25) adapt-

ed to pass only selected wavelengths of light so as to dispose said filter in a light path from said light source, sensor means for detecting the presence and nature of a said filter in said filter receiving means, and interlock means for preventing operation of said light source to carry out photo-treatment except when a said filter appropriate to an intended photo-treatment is present in said receiving means and/or for providing an alarm signal if a said appropriate filter is not present in said receiving means.

19. Apparatus for photo-treatment comprising a light source (13), a filter (25) adapted to pass only selected wavelengths of light disposed in a light path from said light source, said light source being adapted to produce a light flux of at least 250 J/cm²/sec, wherein said filter is a non-interference absorption filter.
20. Apparatus as claimed in Claim 19, further comprising means (38,24,40) for circulating a cooling liquid in contact with said light source.
21. Apparatus as claimed in Claim 20, further comprising a filter (25) adapted to pass only selected wavelengths of light disposed in said light path from said light source and also in contact with said cooling liquid.
22. Apparatus as claimed in any one of the preceding claims, wherein the light source comprises a gas-filled arc lamp.
23. Apparatus as claimed in Claim 22, wherein said gas-filled arc lamp is a xenon or krypton lamp.
24. Apparatus as claimed in any one of the preceding claims, wherein the power supply comprises a capacitor (C), a charging circuit adapted for charging the capacitor to a preselected voltage, a resistor (R) in series between said capacitor and said light source and a discharge switch (IGBT) operable to change from a non-conductive state to a conductive state to cause said capacitor to discharge said light source and back to said non-conductive state again.
25. Apparatus according to Claim 24, wherein the light source is an arc lamp and the power supply comprises a simmer generator adapted for feeding the arc lamp with power at a level which is sufficient to keep the arc in the conductive state.
26. Apparatus as claimed in any preceding claim wherein there is means (62) for focussing the light output to concentrate the same at a selected depth below the surface of the treatment location.

27. A method of treating live tissue for cosmetic purposes, wherein an apparatus as defined in any one of the preceding claims is used.

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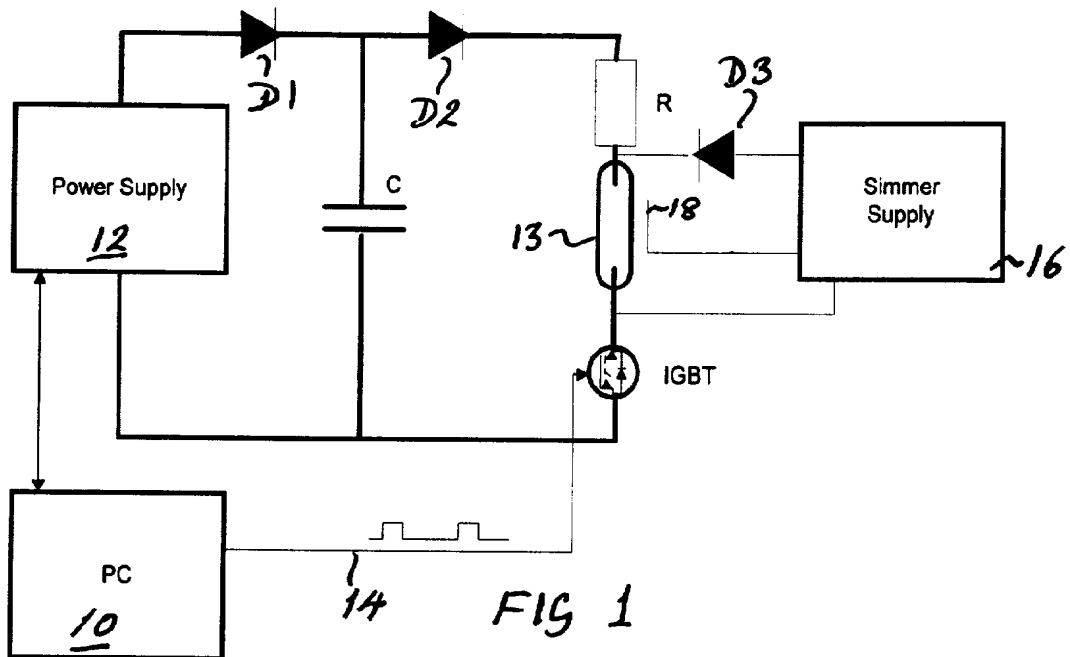
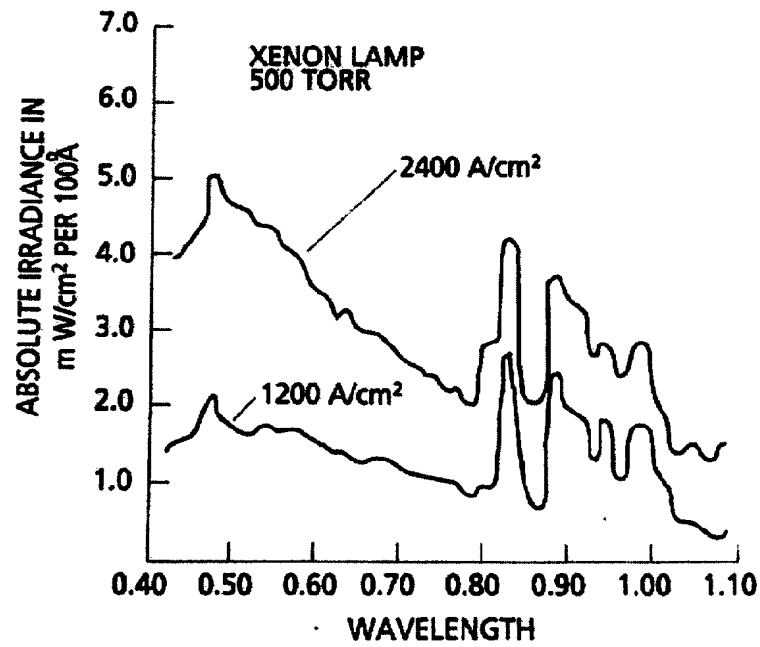
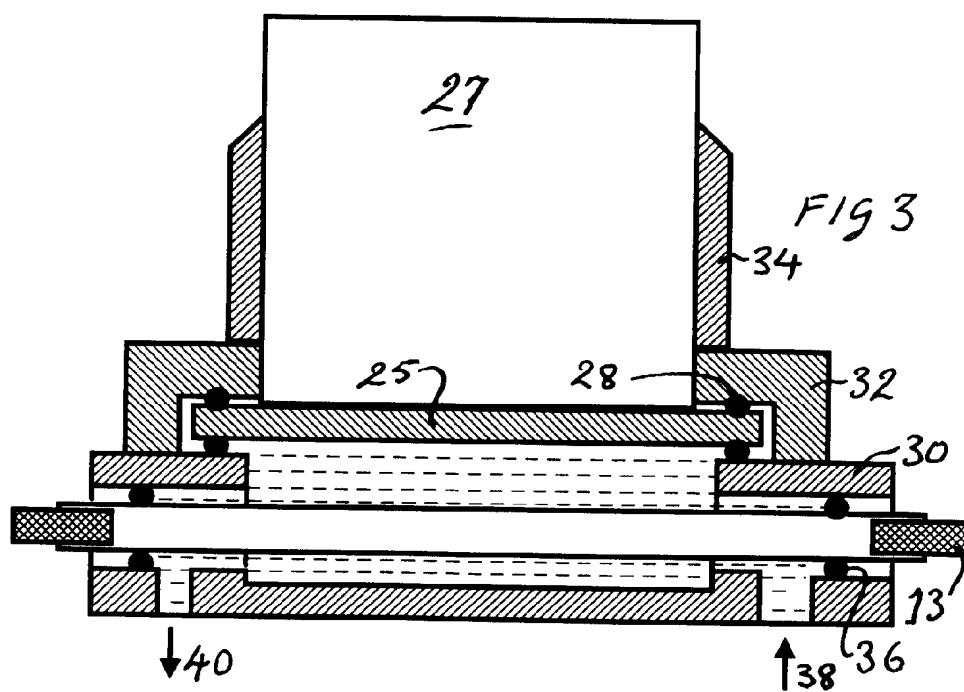
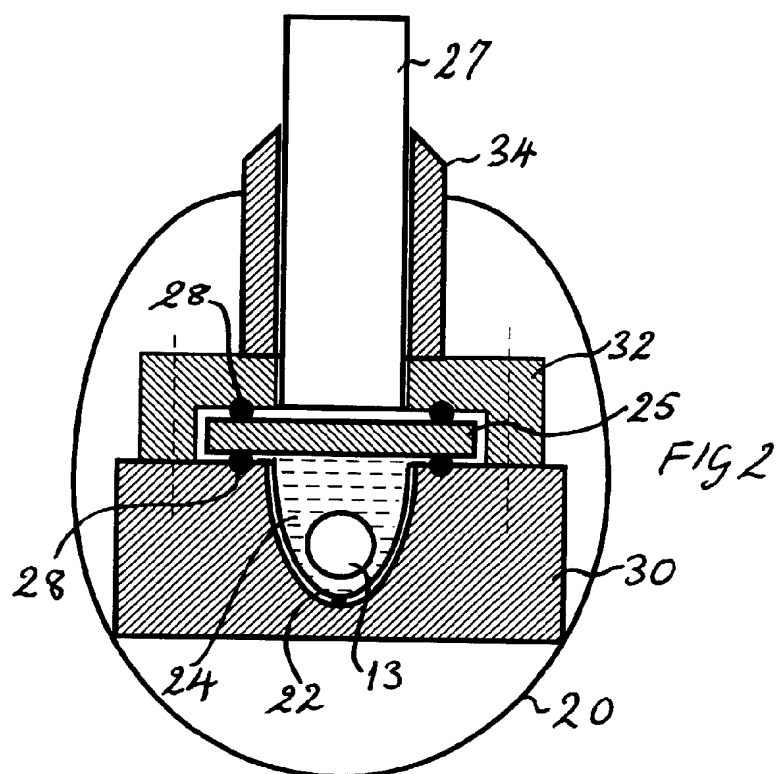
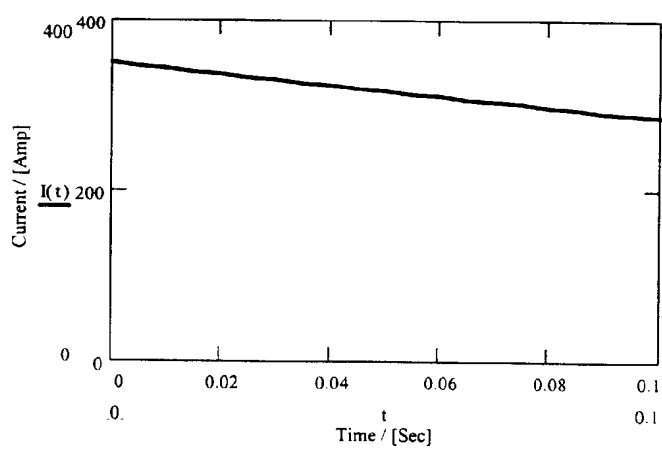
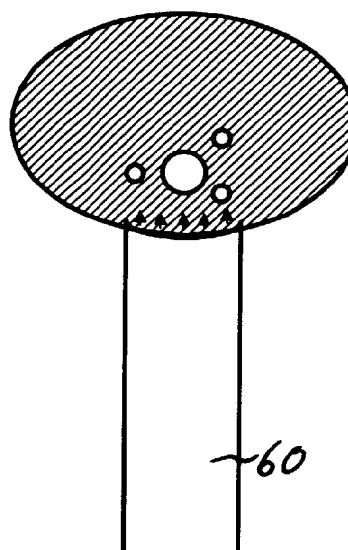
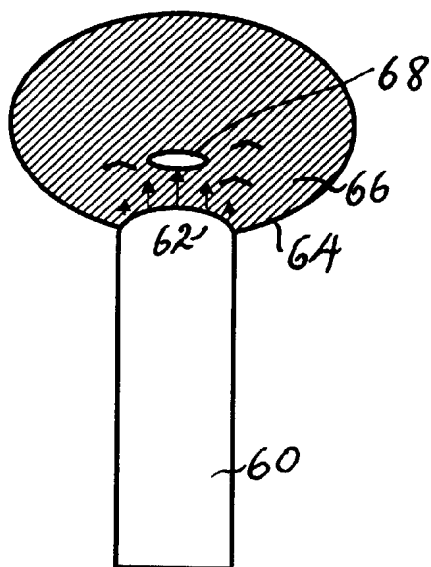


FIG 5



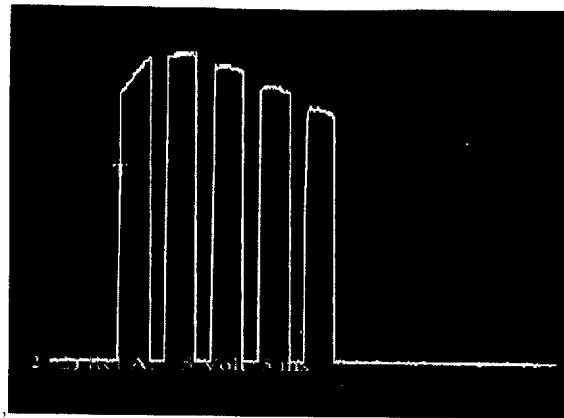




Thin Hair:

5 ms/div , 50A/div

a

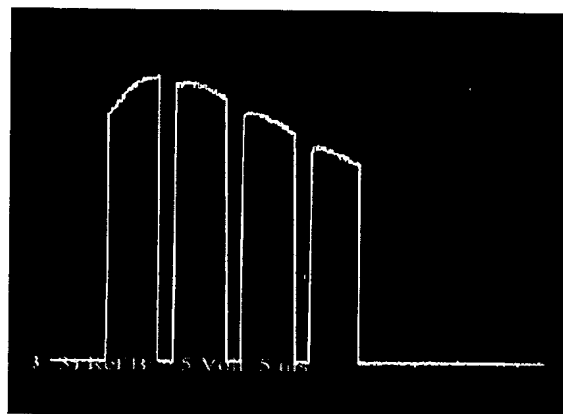


Pulse time = 3 ms, Puls delay = 1,5 ms, N = 5

Normal Hair

5ms/div, 50 A/div

b



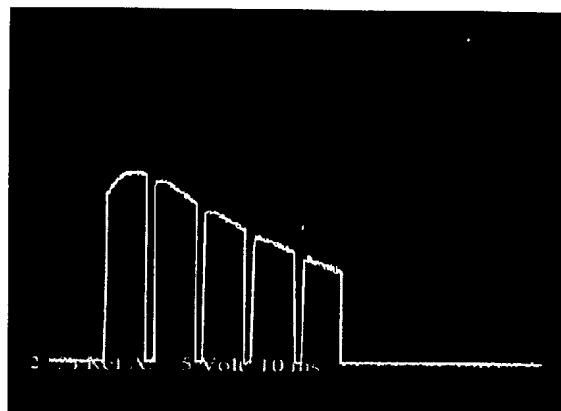
Pulse time = 5ms, Puls delay = 1,5 ms, N = 4

Fig 6

Thick Hair

10 ms/ div, 50 A/div

c



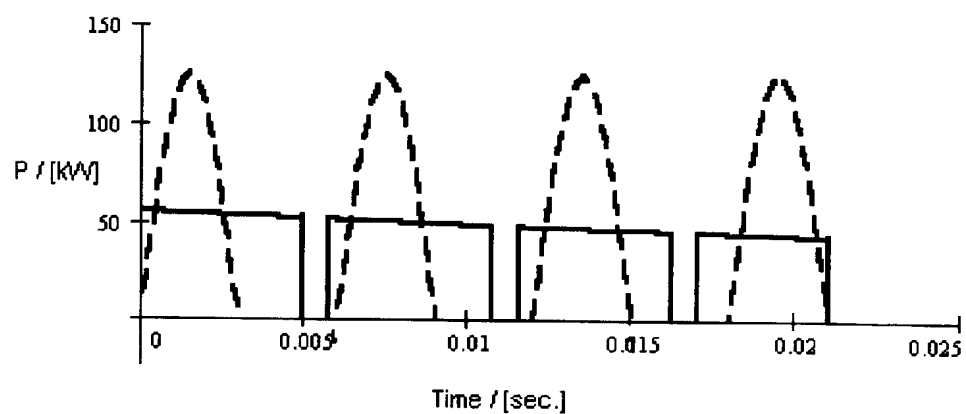


FIG 7

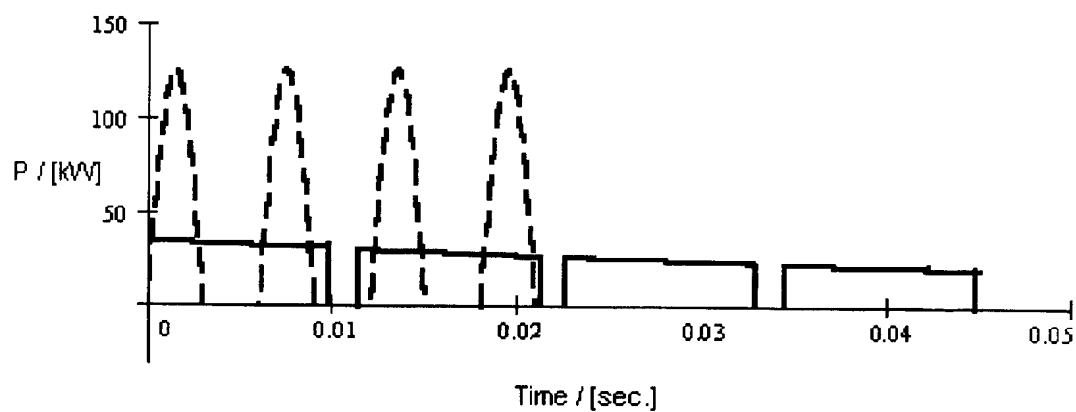


FIG 8

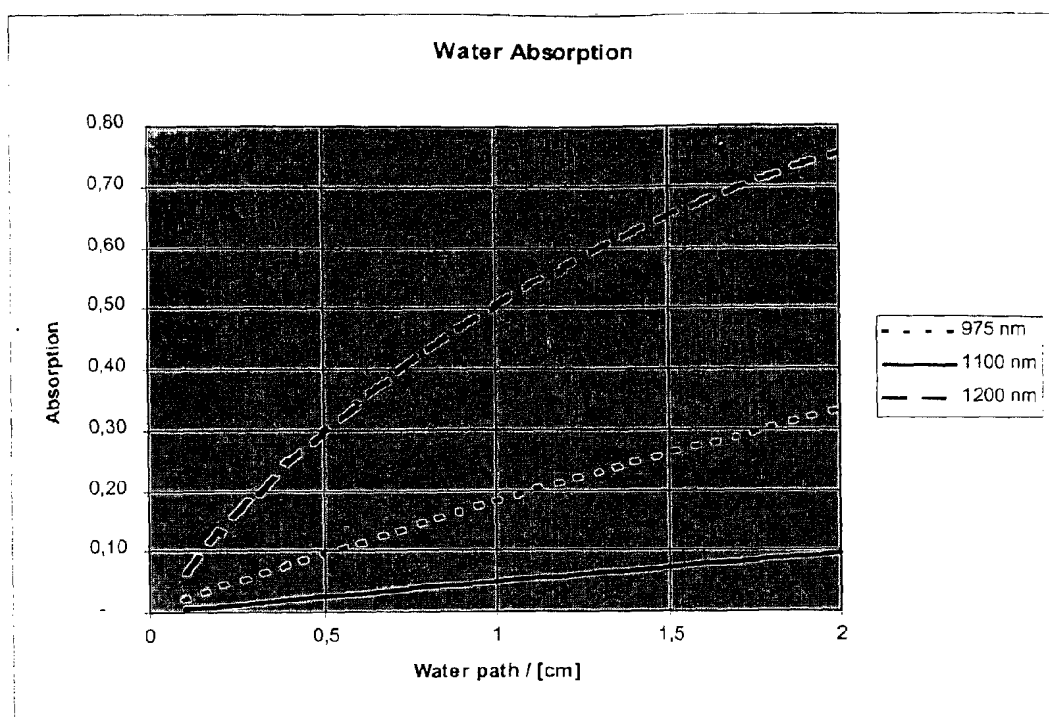
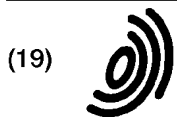


FIG 9



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under INID code 62.

(54) **Small-sized light irradiator for dental use**

(57) A small-sized light irradiator for dental use
comprises

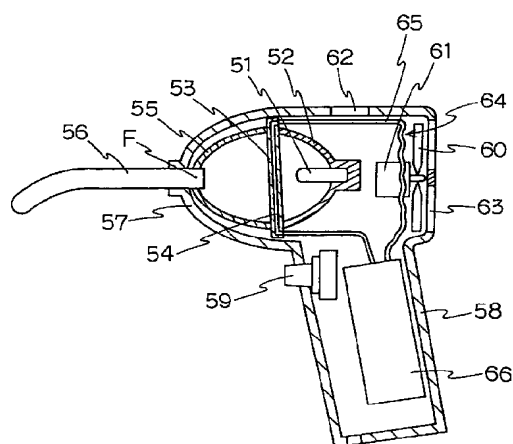
(a) a light source (51) emitting visible, infrared and
ultraviolet light and provided with a substantially
half-ball shaped reflector (52) which converges at
least visible light of said three kinds of light to a pre-
determined direction;

(b) a transparent cell (53, 91) transmissive to visible
light, disposed on the emission side of said light
source (51) to absorb infrared and ultraviolet light,
and having a hollow space therein charged with liq-
uid coolant;

(c) a light guide (56) disposed such that light pass-
ing through said cell (53) impinges on its one side
end and emits from its other side which is directed
to a predetermined direction; and

(d) a casing (57) with a handle (58) enclosing said
light source (51) and transparent cell (53, 91) and
provided with said light guide (56) to its one end.

FIG. 1



EP 0 920 840 A2

Description

[0001] The present invention relates to a small-sized light irradiator for dental use according to the preamble of claim 1, which is used for curing composite resin or curing the inlay within mouth in the dental cure.

[0002] A small-sized light irradiator of this type is known from US 4 623 795 A.

[0003] In the dental curing, it is known that such a method for curing the photopolymerizing resin such as composit resin, which is charged in the dental caries, by irradiation. In such a curing, a small-sized light irradiator has hitherto been used, and known (for example Japanese Examined Utility Model Publication No. 36491/1990).

[0004] In many cases, a halogen lamp is used as a light source of the light irradiator. The halogen lamp outputs visible light, the amount of which is much enough to be suitable for the light source of the light irradiation. However, infrared lights are excessively emitted so as to heat the lamps and neighborhood thereof too much. Conventionally, a filter for the infrared lights is disposed in the output side of the light source and an exhaust fan is disposed in the back side of the light source so that the lamp and the whole device are cooled using the air cooling method. However, such a construction is insufficient for cooling. Therefore, such a halogen lamp as to perform practically cooling effect at low power can be used. It is impossible to cure an inlay in the mouth by photopolymerizing. Further, lights emitted from the halogen lamp includes ultraviolet lights so that there arises such a problem as to cause a suffering due to direct irradiation to the structure of the mouth.

[0005] In the irradiator according to US 4 623 795 A, the light source, reflector and transparent cell are disposed inside the casing in such a manner that a pressurised cooling gas may pass between the inner surface of the casing and the outside of the reflector and transparent cell.

[0006] The object of the present invention is to resolve the above-mentioned problems.

[0007] This object is achieved by a small-sized light irradiator for dental use according to claim 1.

[0008] Advantageous embodiments of the invention are described in the subclaims.

[0009] It is preferred that said transparent cell of visible light is connected to a pump and a heat exchanger via a tube so as to circulate said liquid coolant.

[0010] It is preferred that said transparent cell is composed of a disc-like shaped cell main body having a hollow space therein, and two fine tubes for fluid circulation fixed to said cell main body to communicate with said hollow space.

[0011] Alternatively, said transmission cell of visible light is composed of a conical shaped cell main body having a hollow space therein and two fine tubes for fluid circulation fixed to said cell main body to communicate with said hollow space, wherein said conical

shaped cell is disposed such that a bottom portion of said conical shaped cell is faced to the light source, a tip of said conical shaped cell is faced to a side of one of said light guide.

[0012] It is further preferred that said transparent cell is integrally formed so as to be combined with said reflector, and wherein said reflector has a duplex structure such that a space for communicating with the hollow space within said transparent cell is formed in the reflector.

[0013] It is preferred that a part of said tubes are wound so as to be in contact with an outside surface of said reflector, and wherein said tubes are made of such a metal pipe as to have a good heat conductivity capable of absorbing heat of said reflector.

[0014] It is preferred that said pump and said heat exchanger are accommodated in said casing.

[0015] Alternatively, said pump and said exchanger are arranged in such a manner as to be separated from said casing.

[0016] In the following, preferred embodiments of the invention are described in detail with respect to the accompanying drawings.

FIG. 1 is a sectional view showing one embodiment of the small-sized light irradiator of the present invention;

Fig. 2 is a perspective view showing a transparent cell attached to the light irradiator of Fig. 1;

Fig. 3 is a sectional view taken along with a line III-III in Fig. 2;

Fig. 4 is a sectional view showing another embodiment of the present invention;

Fig. 5 is a partial sectional view showing relevant part in Fig. 4;

Fig. 6 is a sectional view showing an embodiment of the use of the present invention; and

Fig. 7 is a sectional view showing an embodiment of the use of the present invention.

[0017] Fig. 1 shows a small-sized light irradiator, herein numeral 51 indicates light source, for example, halogen lamp emitting visible, infrared and ultraviolet lights. As a light source, xenon lamp, metalhalide lamp and the like can be used other than halogen lamp. Numeral 52 indicates substantially half-ball shaped reflector wherein said halogen lamp and reflect visible light. Numeral 53 indicates transparent cell of visible light which is disposed open space of the reflector 52 i.e. the emission side of the light, charged fluid 54 therein. This all functions as a filter of infrared light absorbent and ultraviolet light. Namely, more than about 80 % of the infrared light and almost all of the ultraviolet light contained in impinging to said cell 53 is absorbed. Numeral 55 indicates another reflector which is disposed outside of the cell 53. Water, ethyleneglycol or silicone oil can be used as fluid 54.

[0018] Numeral 56 indicates light guide formed of

glass fiber, the visible light which transmit the cell 53 impinge from one side thereof, further emit from other side end which is directed to a desired direction. Casing is rotatably provided with light guide 56, and light guide 56 is detachably inserted into the casing. Casing 57 is formed of plastics which is disposed a handle 58. Numeral 59 indicates a power switch which is equipped to the handle 58. Numeral 60 indicates disclosed fan for air exhaust, which is driven by motor 61. Driving fan 60 supplies air from hole 62 for air supply, further evacuate air through hole air exhaust. Adjacent to the fan 60, heat-exchanger 64 which is subjected to blowing air from folded tube 65. Tube 65 is communicated between cell 53 and a pump 66. This pump 66 can be of the construction having a tank for water storage therein (not shown in the figure). With such a construction, fluid 54 charged in the cell 53 is circulated by the pump 66, and cooled at heat exchanger 64. Namely, heated fluid 54 by absorbence of infrared light at the cell 53 is cooled at heat exchanger 64, and again supplied to cell 53. Motor 61 can change its ON-OFF State by operation of power switch 59. Lamp 51, motor 61, pump 66 and power switch 59 are connected by a electronic code which is not shown in the Figure. One motor can be used both as motor 61 and motor for driving of pumps 66. Tubing at the heat exchanger 64 is preferably formed with metal pipe which is superior to heat conductivity such as copper pipe, and at the other portion of tubing elastic silicone tube can be used.

[0019] As shown in Fig. 2 and 3, transparent cell is composed of two transplant glasses 67, 68 which are spaced with a space 69 in 3 mm thickness between them, and fluid 54 is charged to the space 69. Two glasses 67, 68 can be spaced with adhesive 70, to seal space 69. Numeral 71 indicates fine tubes with which communicating space 69, and one of them are used as supply and the other as drain of water. This transparent cell 53 can be composed of heat resistant plastic or heat resistant glass. Two glasses 67, 68 of transparent cell 53 can be also be constructed to concave lens by changing their thickness adequately. Emission light from lamp 51 concave to the spot F, which is indside of the light guide 56 directly, or by reflection at reflector 52 and by further transmission through transparent cell 53.

[0020] Fig. 4 shows another embodiment of the present invention comprising of integrally formed cone-shaped cell main body 78 having vacant space therein by heat resistant glass and two fine tubes 79, 79 for fluid circulation which are communicated with said vacant space thereby fixed to said cell main body 78. In a vacant space of inside of the cell main body 78, aforementioned fluid 54 is charged, Inner surface of the vacant space of cell main body 78 is provided with mirror 90 to reflect light.

[0021] Down-side of the cone of cell main body 78, faces to open end of reflector 52, top of the cone faces to one edge of light guide 56. The light emitted from light source 51 therefore impinge on transparent cell 91, limited almost of infrared light and ultraviolet light by fluid 54 therein, converge to spot F inside of light guide 56 while reflecting to mirror 90, further is introduced by light guide 56. Light impinged to light guide 56 which comprises visible light mainly, is emitted from another side of the guide.

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[0022] Numeral 80 indicates tubular member which is wound on the spherical surface of the reflector 52 in such a manner as to be closely connected each other thereon, made of metal pipe having superior heat conductivity such as copper, one side being connected to fine tube 79 and the other to silicone tube 65, thereby introduced out side of casing 57. As shown in Fig. 5, numeral 81 indicates heat conductive filling plugged to space between metal pipe of cooler 80 and reflector 52 which can be used the solidified powder of aluminium with epoxy resin or silicon resin. Numeral 82 indicates heat exchanger disposed to out side of the casing 57, comprising pump 83, metal pipe 84 having excellent heat conductivity which is coiled to spiral form and fan 85. Pump 83 and metal pipe are connected to tube 65 serially. From fan 85 cool air is blown against metal pipe 84 for the purpose of cooling. Namely, pump 83 makes fluid 54 run out, after passage through transparent cell 91 in the casing 57 via tube 65. Further through cooler 80, then fluid 54 returns to pump 83 tube 65 and metal pipe 84. At transparent cell 91 and cooler 80, the fluid 54 which absorbs infrared light and heated thereby, in cooled down while passing metal pipe 84.

[0023] Fig. 6 and 7 illustrate embodiment of polymerization and curing of composit resin inlay, herein the tip of light guide 56 is provided with small light guide 86. The tip of small light guide 86 is formed into triangle shape. The passed light through inside of small light guide 86 is emitted to outside as it is (shown in Fig. central lower side). Additionally, the light emitted throughout light guide 56, out side of the small light guide 86, impinges to cavitas 87 further disperses to side direction by reflection at the triangle-shaped position. The triangle-shaped portion mentioned above can be of ball like-shaped. After photopolymerizable adhesive is applied to inner siurface of cavitas 87 of teeth 88, composit resin inlay field, and said small light guide 86 is inserted, accordingly light irradiated to composit resin inlay 89 for curing (Fig. 6). Further composit resin inlay 89 is filled into whole cavitas 87, light from light guide 56 is irradiated thereto without small light guide 86. Thus, composit resin inlay 89 can be polymerized and made curing even if it is carried out in a mouth for cavitas 87 with undercut.

[0024] In accordance with the present invention, lights emitted from the light source pass through the transparent cell charged with coolant so that the infrared lights are removed from the emitted lights, and the lights incident to the formed article. The about eighty percent of the infrared lights are cut off, for this reason temperature of the formed article hardly rises and photopolymerization resin used for the pattern is cured. The resin which

is polymerized and cured is taken out from the polymerization reactor so that the next step is readily performed. For this reason, the efficient work can be realized.

[0025] Further, in accordance with the present invention, temperature rise of the formed article is prohibited. For this reason, bad effect on the photopolymerization resin part of the formed article due to the temperature rise such as deformation, boiling of the liquid of monomer or the like is prohibited from generating so that the article having high precision and high strength with bubble free can be obtained.

[0026] Furthermore, in accordance with the present invention, the cooling efficiency is very high. Therefore, the lamp having power higher than that of the conventional one can be used. For this reason, time for polymerization and curing can be shorten so that efficiency of workability can be improved. With respect to the conventional photopolymerization reactor, it takes about three minutes to about four minutes to polymerize, while with respect to the apparatus of the present invention, it can be cured for about 30 to about 40 seconds to cure.

[0027] Further, in accordance with the present invention, the transparent cell is integrally formed of heat resistant glass. Therefore, even if heating and cooling are repeated for long interval in the state as to be equipped near the light source, the coolant is hardly leaked so that durability of the cell can be improved.

[0028] According to the present invention, there is provided a small-sized light irradiation for dental use, wherein the emission light from a light source transmit a cell for transmitting of visible light charged with fluid and because of elimination of infrared light therein and of emission of only visible light through light guide, the emission light can be heatless, accordingly irradiation against cavitas of teeth can be carried out without heating of the effected part. Therefore, the patient can be treated comfortably without pain resulted from heat. Further, according to the invention, there is provided a avoidance of harmful influence against human body by ultraviolet light irradiation, because elimination of ultraviolet light by a cell transmissive to visible light make it possible not to irradiate ultraviolet light against tissue in a mouth.

[0029] Further, according to the invention, there is provided a possibility to use emission lamp with higher power than prior because of extremely high efficiency of cooling of light source, therefore making it possible to carry out curing of inlay in a mouth while difficult because of insufficiency of the light dose in prior art, simultaneously to shorten time for polymerization of photopolymerizable resin. To improve working efficiency. Incidentally, photopolymerizing can be completed in about 30 to 40 sec. by the light irradiator of this invention.

Claims

1. Small-sized light irradiator for dental use, comprising:

- (a) a light source (51) emitting visible, infrared and ultraviolet light and provided with a substantially half-ball shaped reflector (52) which converges at least visible light of said three kinds of light to a predetermined direction;
 - (b) a transparent cell (53, 91) transmissive to visible light, disposed on the emission side of said light source (51) to absorb infrared and ultraviolet
 - light; (c) a light guide (56) disposed such that light passing through said cell (53) impinges on its one side end and emits from its other side end which is directed to a predetermined direction; and
 - (d) a casing (57) with a handle (58) enclosing said light source (51) and transparent cell (53, 91) and provided with said light guide (56) to its one end,
- characterized in that**
- (e) said transparent cell (53, 91) has a hollow space (69) therein charged with liquid coolant (54).

2. Small-sized light irradiator according to claim 1, **characterized** in that said transparent cell (53, 91) of visible light is connected to a pump (83) and a heat exchanger (84) via a tube (65) so as to circulate said liquid coolant.

3. Small-sized light irradiator according to claim 1 or 2, **characterized** in that said transparent cell (53) is composed of a disc-like shaped cell main body (67, 68) having a hollow space (69) therein, and two fine tubes (71) for fluid circulation fixed to said cell main body (67, 68) to communicate with said hollow space (69).

4. Small-sized light irradiator according to claim 1 or 2, **characterized** in that said transmission cell (91) of visible light is composed of a conical shaped cell main body (78) having a hollow space therein, and two fine tubes (79) for fluid circulation fixed to said cell main body (78) to communicate with said hollow space, wherein said conical shaped cell (91) is disposed such that a bottom portion of said conical shaped cell (91) is faced to the light source (51), a tip of said conical shaped cell (91) is faced to a side of one of said light guide (56).

5. Small-sized light irradiator according to one of the preceding claims, **characterized** in that said transparent cell (53, 91) is integrally formed so as to be combined with said reflector (52), and wherein said

reflector (52) has a duplex structure such that a space for communicating with the hollow space within said transparent cell (53, 91) is formed in the reflector (52).

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6. Small-sized light irradiator according to claim 2, **characterized** in that a part (80) of said tubes (65) are wound so as to be in contact with an outside surface of said reflector (52), and wherein said tubes (65) are made of such a metal pipe as to have a good heat conductivity capable of absorbing heat of said reflector (52). 10
7. Small-sized light irradiator according to claim 2 or 6, **characterized** in that said pump (83) and said heat exchanger (84) are accommodated in said casing (57). 15
8. Small-sized light irradiator according to claim 2 or 6, **characterized** in that said pump (83) and said exchanger (84) are arranged in such a manner as to be separated from said casing (57). 20

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FIG. 1

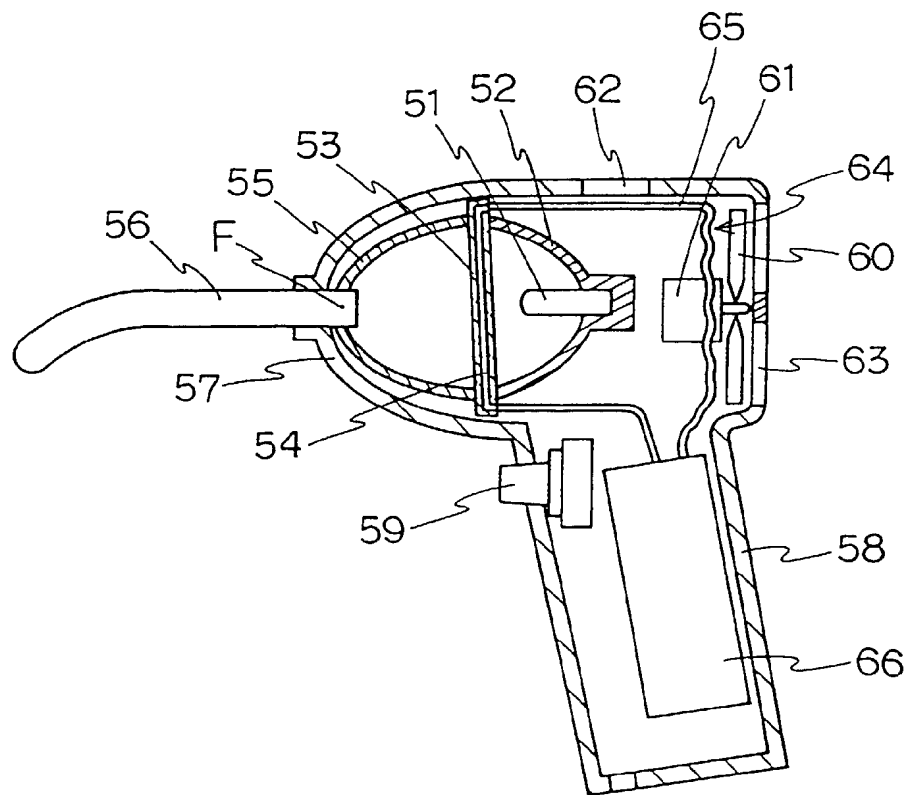


FIG. 2

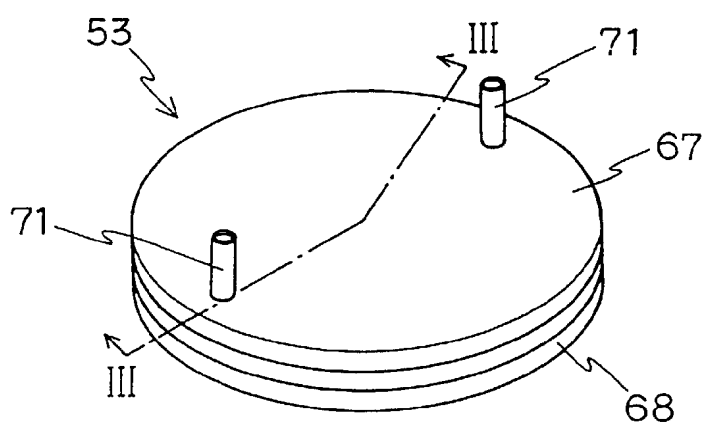


FIG. 3

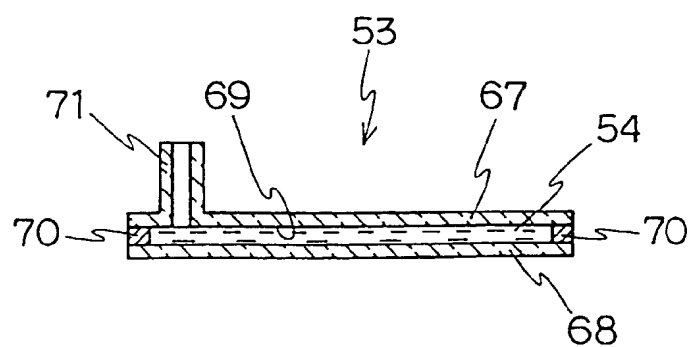


FIG. 4

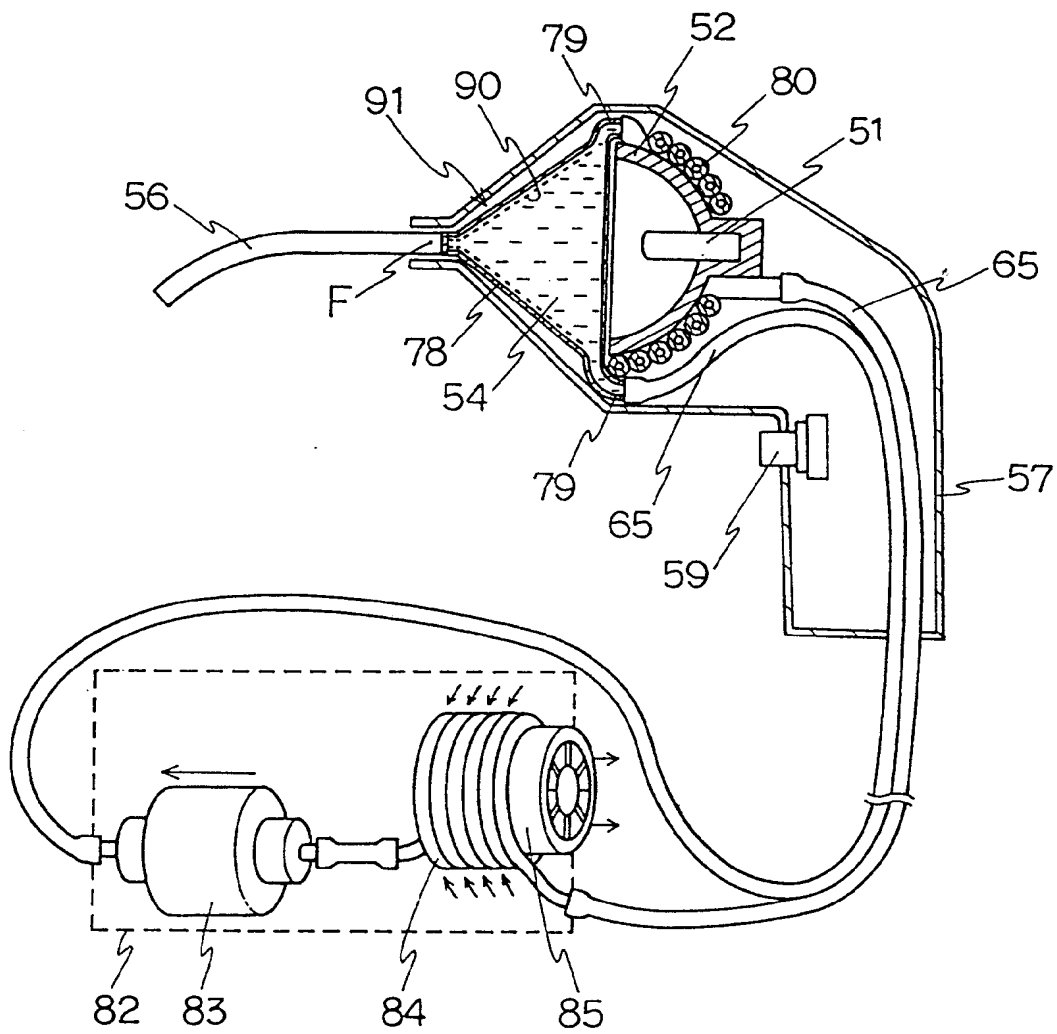


FIG. 5

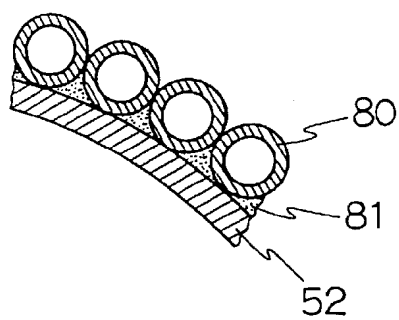


FIG. 6

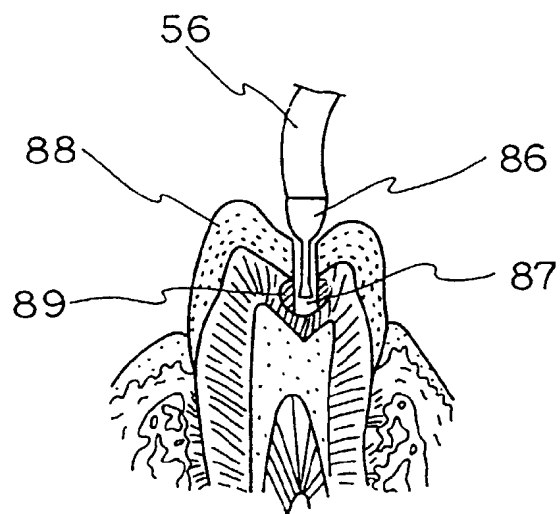


FIG. 7

